

## Experiments with a programmable master hearing aid\*

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**ABSTRACT**—A series of experiments is described tracing the development and application of adaptive paired-comparison testing to the prescriptive fitting of hearing aids. The equipment needed to implement the test procedures became progressively more complex with each new experiment, leading to the development of a digital master hearing aid.

### INTRODUCTION

A particularly useful feature of digital hearing aids is that of programmability. A programmable hearing aid is one in which the electroacoustic characteristics of the instrument can be changed in a pre-determined way within a very short period of time. Instruments of this type can be implemented using either analog or digital components, but it is the digital hearing aid that is particularly well suited for programmable operation.

The capability of being able to switch very rapidly (e.g., within milliseconds) from one set of electroacoustic characteristics to another opens up the possibility of using paired-comparison techniques in the evaluation and prescriptive fitting of hearing aids. The need for an instrument that would allow

adaptive paired-comparison testing of hearing aids played an important role in the development of the first digital master hearing aid.

This paper traces the development and application of adaptive paired-comparison testing to the prescriptive fitting of hearing aids. A series of five experiments is described. Progressively more complex instrumentation was required for each new experiment leading eventually to the development of a digital master hearing aid. Of the five experiments, Experiment 1 has already appeared in print (3,16) and only the results of this experiment are summarized briefly. Portions of Experiment 5 are also in press (25) and, correspondingly, only a brief description is provided of this study.

### EXPERIMENT 1: Adaptive Hearing Aid Prescription Using an Adjustable Analog Hearing Aid

In an early application of adaptive procedures to individualized hearing aid prescription, a wearable master hearing aid was used in which the electroacoustic characteristics of the instrument were systematically adjusted according to the simplicial adaptive procedure (3,16). The hearing aid's characteristics were changed by replacing plug-in modules in the body of the instrument. Traditional analog circuitry was used to modify the average slope, and the lower and upper cut-off frequencies of the frequency gain

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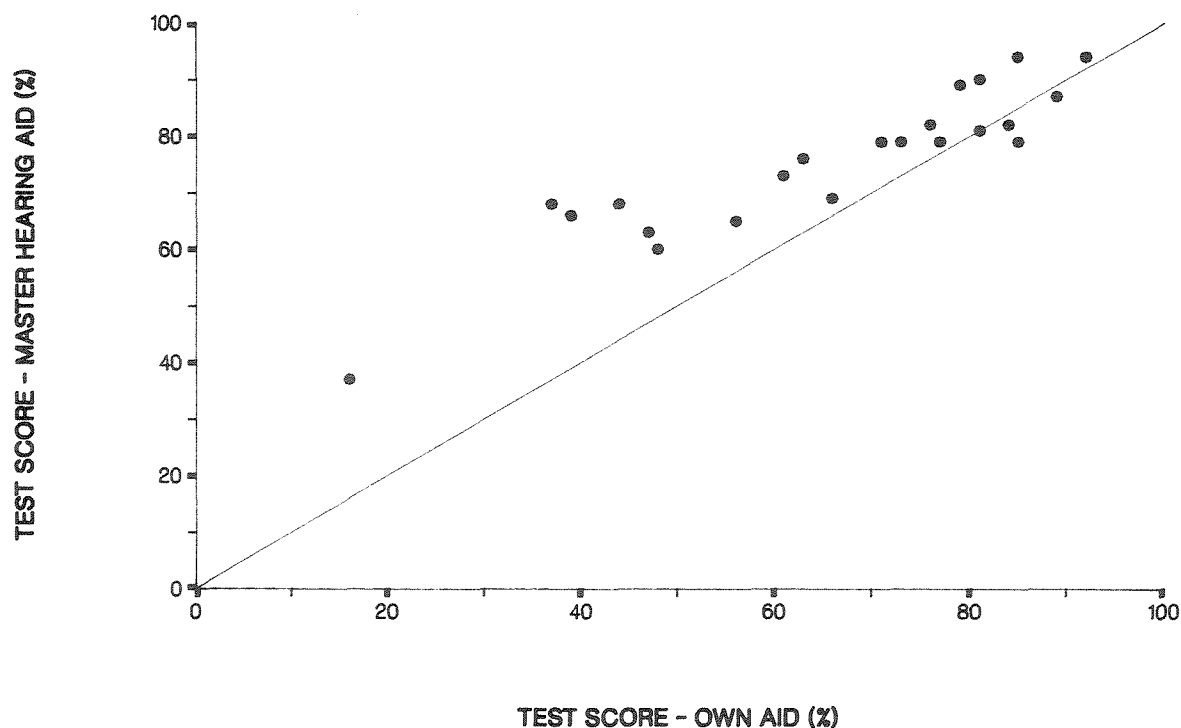


Figure 1.

Comparison of master hearing aid at estimated optimum setting with subject's own aid. Test scores are for the CUNY Nonsense Syllable Test [Diagram reproduced from Collins and Levitt (3)].

characteristic, as well as the maximum power output and form of limiting (e.g., hard or soft limiting).

A standard speech discrimination test was administered after each adjustment to the hearing aid. The score obtained on this test in comparison with the scores on the preceding two trials was used to determine the next adjustment of the hearing aid. A modification of the simplex adaptive procedure was used, see Levitt (11). This method of testing was found to be a time-consuming process and several test sessions of several hours each were required in order to complete the test protocol.

The results of the experiment showed that significant improvements in speech intelligibility could be obtained using this adaptive procedure. The magnitude of the improvements are illustrated in **Figure 1**, which shows speech discrimination scores obtained with the master hearing aid at its estimated optimum setting plotted against speech discrimination scores obtained with the subject's own aid. Any point lying above the diagonal represents improved performance for the master hearing aid. The largest improvements were shown by those subjects having the poorest speech discrimination scores with their

own personal hearing aids. It is not clear whether the subjects with good speech discrimination scores did not show large improvements with the adaptive fitting procedure because there was little room for measuring further improvement using this particular speech discrimination test, or whether these subjects showed little improvement because their own personal hearing aids had been particularly well prescribed to begin with and further improvement would be unlikely using any test instrument.

The important conclusion to be drawn from the above study was that improved performance can be obtained by individualized prescriptive fitting of hearing aids. A second important conclusion was that systematic adaptive procedures could be used to converge on the optimum hearing aid setting using statistically efficient rules, but that the implementation of such procedures was impractical at the time with the technology available and using traditional methods of testing.

The most serious practical problem was the time taken to complete the test protocol (several sessions of several hours each). Other practical problems were human error in adjusting the hearing aid and

in following the rules of a multivariate adaptive procedure. Even the simplified multivariate procedure used in this study was found to be difficult for several of the audiologists implementing the technique.

Additional practical problems related to the specific master hearing aid used in the study in that the number of variables and the ranges of the variables that could be adjusted were limited. For example, adjustment of the frequency gain characteristic was limited to changes in average slope and in upper and lower cut-off frequencies. In many cases, the estimated optimum setting of the hearing aid involved one or more variables being adjusted to the end of their allowable range. In these cases, it was believed that even better results could have been achieved had a larger range of adjustment been available and, more importantly, had there been greater flexibility in terms of which variables could be adjusted.

The results of the above experiments and the belief that even better performance could be obtained using improved instrumentation provided the stimulus for the development of a programmable master hearing aid. For further information on the experiments using a wearable master hearing-aid with plug-in components, see References 2, 3, 11, 16, 17, and 23.

### The Reference Hearing Aid Technique

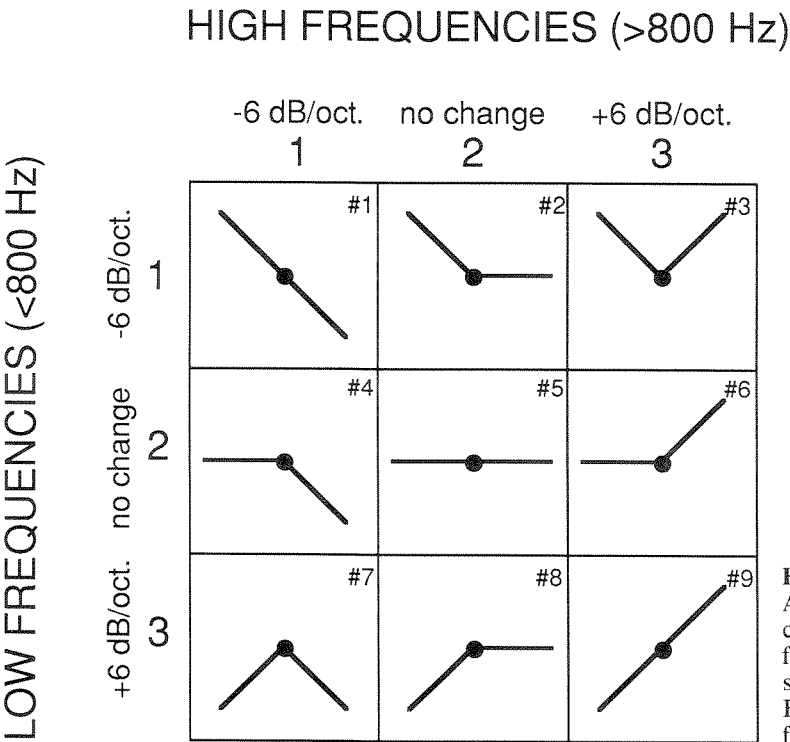
A long-standing problem in hearing aid prescription is that the sound pressure level measured in a standard coupler can be quite different from the sound pressure level at the eardrum. As a consequence, the overall gain of a hearing aid, as predicted from coupler measurements, can differ significantly from the true *in situ* gain. Further, individualized prescription of electroacoustic hearing aid characteristics derived from earphone measurements can be substantially in error if appropriate corrections are not made for differences between earphone and hearing aid receiver in acoustic coupling to the ear and also for the acoustical effects of mounting a microphone on the body (e.g., in or behind the ear).

The reference hearing aid technique was developed to circumvent these problems in individualized prescriptive fitting of hearing aids. Although designed originally for use with master hearing aids (17), it has also been used with conventional hearing

aids (10). The underlying principle of the technique is that the same electroacoustic transducers and method of acoustic coupling used with the hearing aid are also used in obtaining the basic audiological information (e.g., thresholds, discomfort levels) required for the prescriptive fitting procedure. As a consequence, no corrections or adjustments are needed in relating these measurements to the real-ear measurements of the hearing aid. For example, if a measurement of loudness discomfort level shows that X volts at the electrical input to the hearing aid receiver causes discomfort, then it is known that in setting limiting levels for the hearing aid (e.g., maximum power output) the signal level at the receiver should not exceed X volts.

In applying the reference hearing aid technique, it is recognized that the sound transmission path from sound field to eardrum (or more generally, in the case of sound-field testing, from the electrical input to the test loudspeaker(s) to the eardrum) consists of both acoustic and electrical transmission paths. For example, the transmission path from sound field to hearing aid microphone is acoustic, whereas the path from the output of the microphone to the input of the hearing aid receiver is electrical. Since the electrical transmission path can be measured and controlled far more conveniently and more accurately than an acoustic transmission path, all measurements are obtained with the master hearing aid set to a reference condition that is conveniently specified in electrical terms (e.g., a gain of 20 dB with a flat electrical frequency response). This is called the *reference hearing aid* and all audiometric measurements required for the prescriptive fitting procedure are obtained with the reference hearing aid. These measurements can then be related to any convenient point in either the hearing aid or the test system. One very useful reference point is the electrical input to the hearing aid receiver, another is the electrical input to the test loudspeaker. It is also possible to relate any of these measurements to acoustic measurements obtained either in the sound field or in a standard acoustic coupler.

The reference hearing aid technique is, in essence, a generalization of the functional gain procedure (14,28) in which several limitations of the latter technique are avoided. For further discussion of the two techniques, see Reference 12.



**Figure 2.**  
**Adjustment Matrix.** Each row of the matrix shows the change in slope of the frequency-gain characteristic for frequencies below 800Hz. (E.g., for Row 1 the change in slope is -6 dB/octave and for Row 3 it is +6 dB/octave.) Each column of the matrix shows the change in slope for frequencies above 800Hz.

**EXPERIMENT 2: Prescriptive Fitting of a Monaural Hearing Aid Using a Paired-Comparison Technique**

**Background**

The time taken for a conventional speech discrimination test is on the order of several minutes. This time is far too large, by at least an order of magnitude, to be of practical use in an adaptive prescriptive fitting procedure. In contrast, the time taken for a paired-comparison judgement of relative intelligibility is on the order of tens of seconds. Further, paired-comparison judgements can also be obtained of relative sound quality, or of any other subjective dimension likely to be of importance in hearing aid prescription. A problem with the paired-comparison technique is that the judgements are inherently subjective, whereas a speech-discrimination score is a relatively objective measure. It was thus of interest to compare hearing aid prescriptions obtained by means of a paired-comparison technique with prescriptions obtained using speech discrimination testing.

An experiment was performed in which a hearing aid was prescribed using a round-robin procedure.

In this procedure, an initial estimate of the optimum hearing aid for a given subject is obtained and then compared systematically with hearing aids that differ along one or more dimensions from this initial estimate. In this particular implementation of the round-robin procedures the slope of the frequency-gain characteristic was adjusted independently in the low and high frequency regions.

**Figure 2** shows a 3 × 3 matrix forming a set of nine hearing aids. Each cell in the matrix corresponds to a different hearing aid, the frequency-gain characteristic of which is equal to the initial estimate of the frequency-gain characteristic modified by the adjustment shown in the corresponding cell of the matrix.

Each row of the matrix corresponds to a 6 dB/octave adjustment in the slope of the frequency-gain characteristic below 800 Hz; i.e., for all cells in Row 1, the slope of the frequency-gain characteristic is reduced by -6 dB/octave below 800 Hz, for Row 2 there is no change in slope below 800 Hz, and for Row 3 the slope is increased by +6 dB/octave below 800 Hz.

The columns of the matrix correspond to 6 dB/octave increments in the slope of the frequency-gain characteristic above 800 Hz. Column 1 corre-

sponds to a  $-6$  dB/octave decrement in slope above 800 Hz; Column 2 corresponds to no change in slope; and, Column 3 corresponds to a  $+6$  dB/octave increment. Note that the frequency-gain characteristic corresponding to the central cell (2, 2) is equal to the initial estimate since there is no change in slope in either the low or high frequencies. The frequency of 800 Hz was chosen as the dividing frequency between low and high frequencies since it is the center frequency of the one-third-octave band midway in the frequency range of interest (100 Hz to 6000 Hz). The upper frequency bound was set by the bandwidth of the hearing aid receiver and is typical of most good quality commercially available hearing aids.

The hearing aid corresponding to each cell of the matrix is compared with the hearing aids corresponding to every other cell in the matrix. This set of comparisons is referred to as a round-robin tournament. The hearing aids can be ranked in terms of their speech discrimination scores or, for paired-comparison judgements, the hearing aids can be ranked according to the number of times each hearing aid is chosen over the other hearing aids in the round-robin tournament. The "winning" cell in the round-robin tournament becomes the second estimate of the optimum frequency-gain characteristic for the subject. A third estimate can be obtained by forming a new matrix using the winning cell of the first tournament as the central cell of the next tournament. In an adaptive fitting procedure based on the round-robin technique, the tournaments are repeated iteratively until successive estimates of the optimum hearing aid show no further improvement. Because of the time taken for each tournament, it was considered sufficient for the purposes of this study to terminate the adaptive procedure after the second estimate was obtained.

### Instrumentation

**Figure 3** shows a block diagram of the master hearing aid used for this study. Two sets of pre-programmed filters (F1 through F9) were used to implement the adjustments to the frequency-gain characteristic shown in **Figure 2**. Two sets of filters were used so that the subject could switch instantaneously, by means of Switch S1, from one hearing aid condition to another. Filter F0 was a General Radio one-third-octave band multifilter (Model 1925) used to obtain the initial estimate of the optimum

frequency-gain characteristic. This filter was adjusted manually for each subject at the start of each test session.

The subject sat in a sound-treated audiometric test room. The location of the subject's seat was fixed near the center of the room for all experiments. Two loudspeakers were used, one directly in front of the subject at a distance of 4 feet. The second loudspeaker was opposite to the ear fitted with the hearing aid, at a distance of 3 feet. Speech signals and one-third-octave band test signals were presented at 0 degrees azimuth. Competing signals were presented at 90 degrees azimuth.

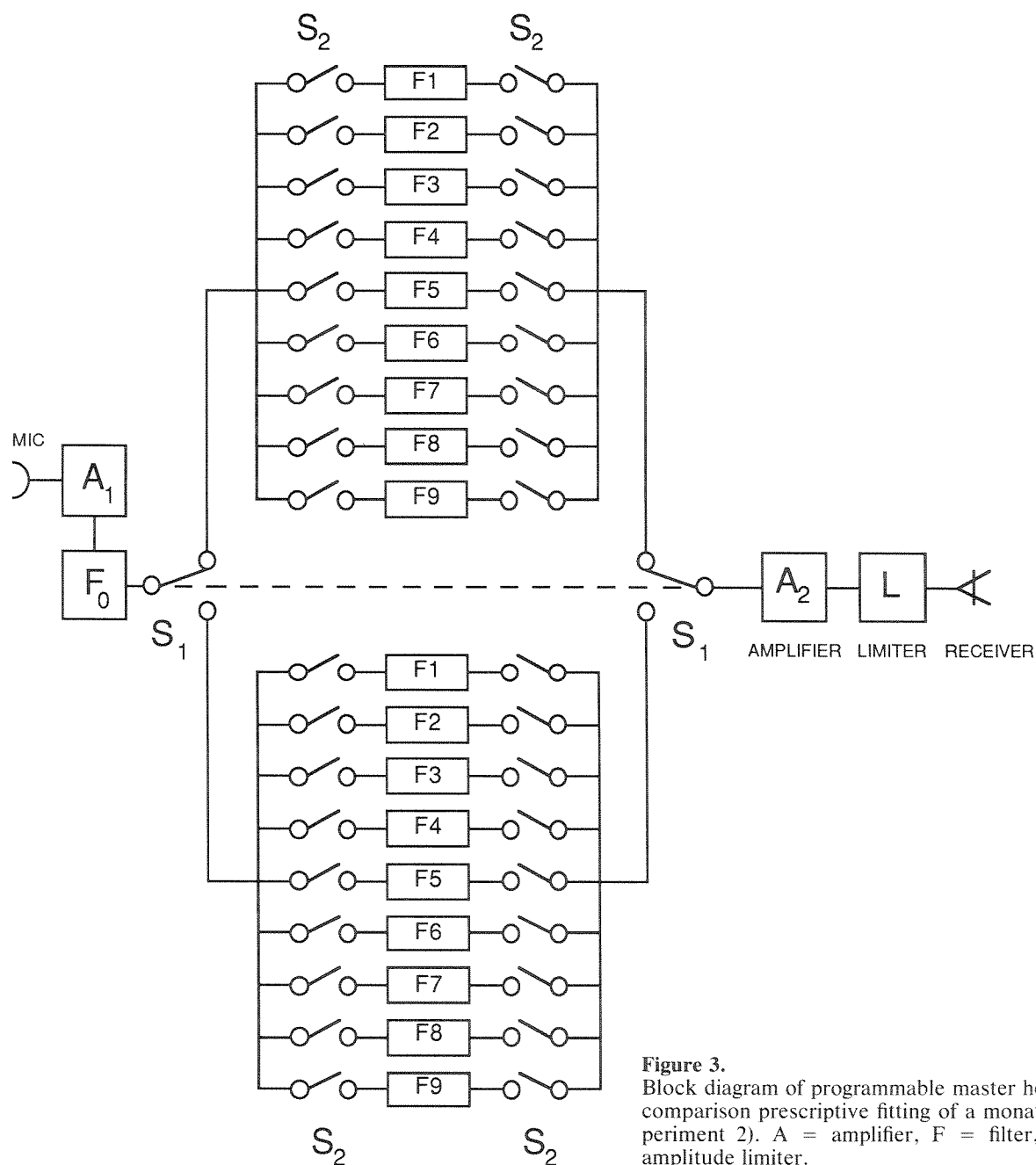
The subject wore a conventional post-auricular hearing aid case fitted with a standard hearing aid microphone. The output of the microphone was led to a pre-amplifier mounted in a body worn unit. This unit also contained a power amplifier and amplitude clipper. The output of the power amplifier was led to a Danavox button-type hearing aid receiver mounted in a custom-made shell earmold. The clipper was set to a maximum level of 125 dB SPL and was used to protect the subject from potentially hazardous intense stimuli.

The multifilter was used to shape the long-term average speech spectrum to be parallel to the subject's loudness discomfort level (LDL). The nine pre-programmed filters were used to modify the frequency-gain characteristic of the master hearing aid, as indicated in **Figure 2**. Two channels were used to allow the subject to make instantaneous comparisons between pairs of frequency-gain settings. The subject could select either channel, during the paired-comparison phase of the testing, by means of a hand-held remote control toggle switch.

The dynamic range of the test system was roughly 50 dB. This presented a slight problem for the experimental condition requiring a  $+6$  dB/octave increase in both the low and high frequencies. The frequency range from 100 to 6000 Hz covers 6 octaves which, if added to the high-frequency boost typically required for the initial estimate of the optimum frequency-gain characteristic, resulted in a dynamic range approaching the maximum available.

### Subjects

Five men and five women with sensorineural hearing loss served as subjects for this study. Subjects ranged in age from 20 to 65 years of age with



**Figure 3.** Block diagram of programmable master hearing aid for paired-comparison prescriptive fitting of a monaural hearing aid (Experiment 2). A = amplifier, F = filter, S = switch, L = amplitude limiter.

hearing losses that varied from moderate to severe. Bone-conduction thresholds were within 10 dB of air-conduction thresholds at all frequencies for all subjects. Mean speech recognition ability as measured with the CID W-22 recordings was 67 percent with a standard deviation of 18.2 percent.

Subjects differed with regard to their experience in wearing amplification. Years of hearing aid use ranged from 3 to 23 years. One subject had rejected

the use of a hearing aid after a 30-day trial period. Eight of the subjects wore their hearing aids during all waking hours, and one wore his aid only occasionally.

#### Test Procedures

The experimental protocol consisted of six 2-hour sessions. Subjects were allowed to schedule their visits at their convenience, but were encouraged to

return at weekly intervals. Audiological data and earmold impressions were obtained during each subject's initial test session. Custom earmolds were made for all subjects, as all further testing would be done in the aided condition.

The initial estimate of the optimum frequency-gain characteristic was obtained using the LDL-spectrum-shaping technique (1). The master hearing aid was set to the reference setting (nominally flat response), and aided loudness discomfort levels were obtained for one-third-octave bands of noise presented at 0 degrees azimuth in the sound-field. The nontest ear was occluded with a custom earmold without the sound bore.

LDL measurements were obtained using the instructions proposed by Morgan, Wilson, and Dirk (24). A simple up-down adaptive procedure with 2 dB step intervals was used to approximate the 50 percent point on the psychometric function (11). The noise bands were pulsed with an on-off time of 250 ms., and a rise-decay time of 25 ms. Three pulses were presented at each step interval.

It was not possible to obtain loudness discomfort levels for one-third-octave bands below 200 Hz. This was due, in part, to the normal statistical fluctuations encountered with narrowband random signals. In addition, several of the subjects appeared to require signal levels above the safety level of 125 dB SPL at the extremes of the frequency range being tested. In these instances, the maximum allowable signal level (125 dB SPL) was used.

Previous research on the LDL-spectrum-shaping technique showed an upward spread of masking effect when the low frequency speech spectrum was amplified close to the loudness discomfort level (1). For this reason, the first estimate of the optimum frequency-gain characteristic was obtained by shaping the speech spectrum to lie parallel to the subject's loudness discomfort level curve, and then adding a 6 dB/octave roll-off for frequencies below 800 Hz. The multifilter was used to provide the necessary frequency shaping in one-third-octave band steps, using the LDL data obtained for the one-third-octave bands of noise in the sound-field. The settings of the multifilter were determined individually for each subject such that, when the speech stimuli used in subsequent testing was presented over the loudspeaker, each one-third-octave band of the speech signal was a fixed number of decibels below the subject's loudness discomfort level. The additional

6 dB/octave roll-off below 800 Hz was then added to the multifilter settings.

The overall level of the speech signal was adjusted separately for each subject. This was done by having the subject adjust the gain of the master hearing aid until the speech sounded maximally intelligible. Gain levels for maximum intelligibility were determined individually for each cell in the adjustment matrix (Figure 2). Subjects listened to a recording of continuous speech presented at a fixed level (72 dB SPL at the position of the listener's head), and were instructed to adjust the gain of the master hearing aid to make the speech most intelligible. The instructions used were adapted from Cox (4).

Speech discrimination tests were administered for each of the nine cells shown in Figure 2. The sequence of testing was randomized. For the round-robin tournament, paired-comparison judgments of relative intelligibility were obtained for all possible pairs of cells in the matrix. As before, order of testing was randomized.

### Test Materials

Speech discrimination was measured using the CUNY Nonsense Syllable Test (21). This test has been demonstrated to be sensitive to the consonant confusions made by the hearing impaired listener (17) and can be administered repeatedly with negligible learning effects (6). Six protocols of the original test were selected and modified slightly for use in this study. Each protocol consisted of nine subtests.

The paired-comparison judgments of relative intelligibility were made using a recording of continuous speech. Subjects listened to a 1.7-minute recorded passage of continuous speech made by the same talker who recorded the CUNY Nonsense Syllable Test (NST). The talker was instructed to control intensity variations because a passage of relatively constant level was required. A graphic level recording revealed that the long-term rms level of the speech remained constant within 2 dB. In making the paired-comparison judgments, subjects were instructed to switch freely between aids "A" or "B," as marked on the remote control switch, and select that aid that made the speech more intelligible. Instructions for the paired-comparison procedure were based on those developed by Studebaker (33).

## Results

The speech discrimination scores obtained for each subject for each of the nine cells in the adjustment matrix are shown in **Table 1**. Also shown are mean scores and standard deviations for each subject across conditions. The last three rows of the table show the mean score for each subject, the observed standard deviation, and the expected standard deviation, respectively. The expected standard deviation was obtained by assuming a binomial sampling distribution about the mean score. This is a conservative assumption. Dubno et al. (6), for example, found the distribution of scores for the CUNY Nonsense Syllable Test to be slightly greater than that predicted by the binomial assumption. For only 2 of the subjects was the measured standard deviation of the test scores across conditions significantly greater than the expected standard deviation, indicating no significant effect of the nine test conditions.

**Table 2** shows the rankings of the nine hearing aids for each subject. The upper half of the table shows the rankings obtained by comparing speech-discrimination scores. The lower half of the table shows rankings obtained from the paired-comparison judgments in the round-robin tournament. The bottom row of the table shows the Spearman rank correlation coefficient for the two sets of ranks. Three of the subjects (A, D, and E) showed a statistically significant correlation ( $p < 0.05$ ) between the two sets of ranks.

## Discussion

The most striking result was that the speech-discrimination scores did not differentiate well between the nine hearing aids. For 8 of the 10 subjects, the standard deviation between conditions was no greater than the expected standard deviation due to random sampling effects, as predicted by binomial sampling theory. This estimate of the expected standard deviation also does not take into account other sources of random variation, and is slightly less than the test-retest variability obtained empirically (6).

One of the reasons for the lack of statistically significant differences among test conditions was that the subject was allowed to adjust the output level for each condition, as would be the case in everyday listening with hearing aids. Thus, although there were large differences among the frequency-gain characteristics for the nine test conditions, the adjustment of the hearing aid output to the most intelligible level for each subject helped reduce the effect of these differences on measured intelligibility. A similar result has recently been obtained by Sullivan et al. (36) in comparing different predictive formulae for prescribing hearing aids. The above findings are also consistent with those of Shore, Bilger, and Hirsh (30), who concluded that traditional methods of speech audiometry are not sufficiently sensitive to differentiate reliably between hearing aids.

The paired-comparison technique, in contrast was

**Table 1.**  
Speech Discrimination Scores for the Hearing Aids Corresponding to the Nine-Cell Matrix.

Hearing Aid #	Cell (row #, col #)	Subjects									
		A	B	C	D	E	F	G	H	I	J
1	(1,1)	63.9	50.0	27.4	58.2	75.3	63.7	39.7	61.0	44.5	48.6
2	(1,2)	66.4	50.0	38.3	45.2	77.4	63.7	34.2	54.8	49.3	51.4
3	(1,3)	67.1	41.8	38.3	63.0	65.8	66.4	41.1	61.7	59.0	52.1
4	(2,1)	67.8	39.0	26.7	67.1	73.3	71.2	41.1	65.8	45.2	59.0
5	(2,2)	58.2	45.2	41.1	55.5	69.9	69.9	40.4	69.9	46.6	49.3
6	(2,3)	59.6	52.1	44.5	73.3	65.1	68.5	43.8	53.4	56.2	60.3
7	(3,1)	64.4	45.2	30.1	56.2	73.3	63.0	43.8	58.2	41.1	51.4
8	(3,2)	50.7	45.9	48.6	60.3	65.8	65.1	38.4	58.9	50.7	47.9
9	(3,3)	54.1	47.3	37.7	49.3	71.9	75.3	41.1	52.1	39.0	55.5
Mean		61.4	46.3	37.0	58.7	70.9	67.4	40.4	59.5	48.0	52.8
Std. deviation		6.1	4.2	7.6*	8.6*	4.5	4.1	2.9	5.8	6.6	4.5
Expected std. dev.		5.6	5.7	5.5	5.6	5.2	5.3	5.6	5.6	5.7	5.7

\* = significantly greater than expected standard deviation ( $p < 0.1$ )



**Table 2.**  
**Ranking of Hearing Aids**

Rank order as determined by speech discrimination score. (1 = best score.)										
Subject	A	B	C	D	E	F	G	H	I	J
Hearing Aid										
1	5	2.5	8	5	2	7.5	7	4	7	8
2	3	2.5	4.5	9	1	7.5	9	7	4	5.5
3	2	8	4.5	3	7.5	5	4	3	1	4
4	1	9	9	2	3.5	2	4	2	6	2
5	7	6.5	3	7	6	3	6	1	5	7
6	6	1	2	1	9	4	1.5	8	2	1
7	4	6.5	7	6	3.5	9	1.5	6	8	5.5
8	9	5	1	4	7.5	6	8	5	3	9
9	8	4	6	8	5	1	4	9	9	3

Rank order as determined by paired comparison judgments. (1 = most frequently selected hearing aid.)										
Subject	A	B	C	D	E	F	G	H	I	J
Hearing Aid										
1	2	2.5	8	8	1	6	9	1	5.5	2.5
2	2	2.5	6	6.5	2	6	5	3	1	6.5
3	6	2.5	2	4	5	2	1.5	5	3.5	5
4	2	2.5	7	1.5	3	8	7	2	3.5	2.5
5	4	6	4	6.5	5	3.5	3.5	7	2	1
6	8	8	3	4	8	6	6	6	5.5	8
7	5	5	9	1.5	5	9	3.5	4	8	6.5
8	7	7	5	4	7	1	1.5	8	7	4
9	9	9	1	9	9	3.5	8	9	9	9

Spearman rank correlation coefficients.										
Subject	A	B	C	D	E	F	G	H	I	J
r =	0.62*	-0.28	0.50	0.61*	0.76*	0.29	-.03	0.32	0.49	-0.49

\* = statistically significant ( $p < 0.05$ )

not only much quicker to administer (each judgment requiring tens of seconds as opposed to minutes for a speech discrimination test), but the data exhibited much less variability. Subjects not only were found to be consistent in their judgments but, with one exception (Subject B), there were relatively few circularities in the paired-comparison judgments leading to ties in the rankings. Circularities in paired-comparison judgments refer to situations of the type where condition X is selected over Y, which is selected over Z, which, in turn, is selected over X (i.e.,  $X > Y > Z > X$  forms a circularity). Circularities can occur either as a result of random guessing or because of consistent differences in judgmental criteria between conditions.

The correlation between the rankings of the nine test hearing aids obtained by speech-discrimination

scores and by paired-comparison judgments was low. Only three of the subjects showed a statistically significant correlation between the two sets of rankings. The low correlations appeared to be due largely to the high test-retest variability of the speech-discrimination scores. On the average, those subjects showing a higher range of speech discrimination scores between test conditions (relative to the test-retest variability of these scores) also showed a higher correlation between the two sets of rankings. Low correlations are to be expected when one or both sets of measurements are subject to high test-retest variability. It is possible to estimate the average reduction in the measured correlations if the test-retest variability is known (7). Using the expected standard deviation of the speech-discrimination scores due to random sampling as a guide, it is estimated

that the measured correlations will, on the average, be equal to 0.68 of the true correlations without random error.

Speech discrimination scores have traditionally been used as the primary test instrument in laboratory evaluations of hearing aids. In practice, however, hearing aid selection depends very heavily on the purchaser's preferences. The paired-comparison procedure provides a much closer approximation to the latter process. While speech discrimination tests provide a more objective measure than paired-comparison judgments (of relative intelligibility), the high test-retest variability of the former measure reduces its practical usefulness considerably.

It is clear from the above that, although speech discrimination testing is of great value in research studies, it is of limited practical value in the prescriptive fitting of hearing aids. Further, there is no *apriori* reason for assuming that intelligibility is the primary criterion measure in hearing aid prescription. Other criteria, such as quality of sound reproduction, convenience of use, and long-term comfort, are also important considerations in hearing aid selection and, for many clients, may well exceed the importance of relative intelligibility as the primary criterion.

In summary, the paired-comparison technique was found to be considerably more efficient than the use of traditional speech discrimination tests in the prescriptive fitting of hearing aids. Because of the high variability shown by the speech discrimination tests, only very little could be said about the consistency of the two procedures. In order to implement the paired-comparison technique, however, it is necessary to have a programmable master hearing aid in which the subject can switch rapidly between paired sets of hearing aid conditions.

### **EXPERIMENT 3: Binaural Hearing Aid Prescription Using A Paired-Comparison Technique**

#### **Introduction**

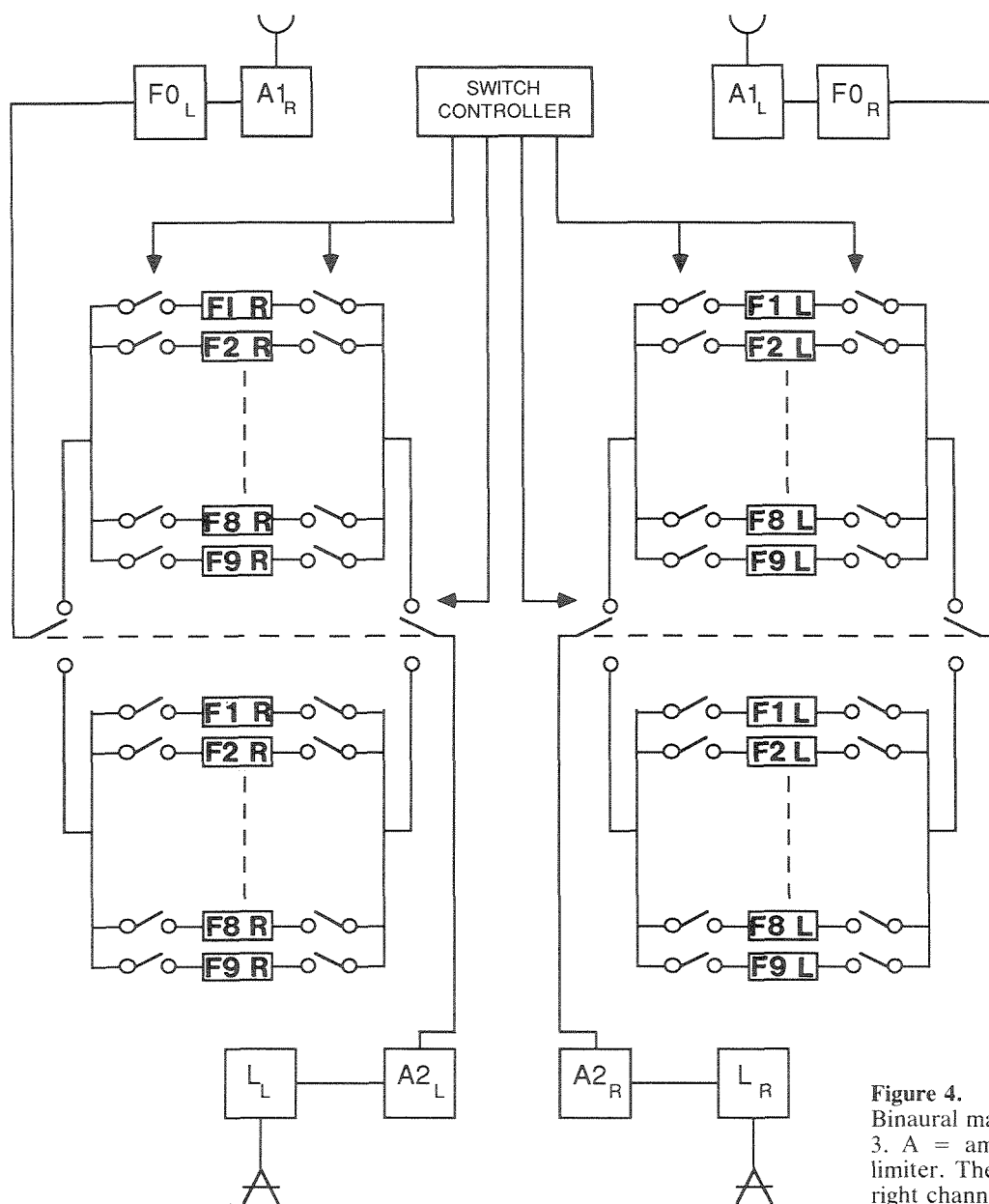
Given the great efficiency of the paired-comparison technique, it was decided to investigate the use of that technique in the prescriptive fitting of a binaural hearing aid. The number of variables to be considered in binaural hearing aid prescription is more than twice that of monaural hearing aid prescription. Given the multivariate nature of the problem, a doubling of the number of variables raises

formidable difficulties. There have been few attempts at developing and evaluating prescriptive fitting procedures for binaural hearing aids, presumably because of these difficulties. It was believed that the relatively high efficiency of the paired-comparison procedures would at least provide a means for tackling this difficult problem.

In order to implement the paired-comparison techniques for binaural applications, a programmable 2-channel master hearing aid was constructed. A block diagram of the system is shown in **Figure 4**. The binaural master hearing aid differed from its monaural counterpart in two important respects. First, the binaural aid consisted of two independent channels, each channel being the equivalent of the monaural master hearing aid shown in **Figure 3**. Second, in order to facilitate the many switching operations required for the paired-comparison technique, relay controlled switches were installed. This represented an early step towards a more flexible computer-controlled switching system that, in turn, led to the development of a fully computerized hearing aid.

Another important difference between the monaural and binaural systems was that the pre-programmed filters F1 through F9 were not adjusted for changes in the slope of the frequency response, but were adjusted to provide a stepped response of the form shown in **Figure 5**. This was done because the extreme slope conditions taxed the limited dynamic range of the system, as discussed in the preceding section. The size of the steps used with the stepped frequency response was determined from the subject's judgments of relative loudness as a function of signal level in each band.

A relatively simple approach to the prescriptive fitting of binaural hearing aids is that in which the aid for each ear is fitted independently; i.e., each ear is fitted with the best hearing aid for that ear independently of the signals reaching the opposite ear. This seems to be the approach commonly used for persons with a symmetrical binaural hearing loss. An experiment was thus performed to determine whether or not the best frequency-gain characteristic for each ear monaurally, when combined, would yield the best binaural fitting. In order to test this hypothesis, alternative combinations of hearing aids were also evaluated. These included frequency-gain characteristics that would not necessarily be chosen for monaural fittings, but might optimize speech intelligibility binaurally (9,29).



**Figure 4.**  
Binaural master hearing aid used in Experiment 3. A = amplifier, F = filter, L = amplitude limiter. The suffixes L and R indicate left and right channels, respectively.

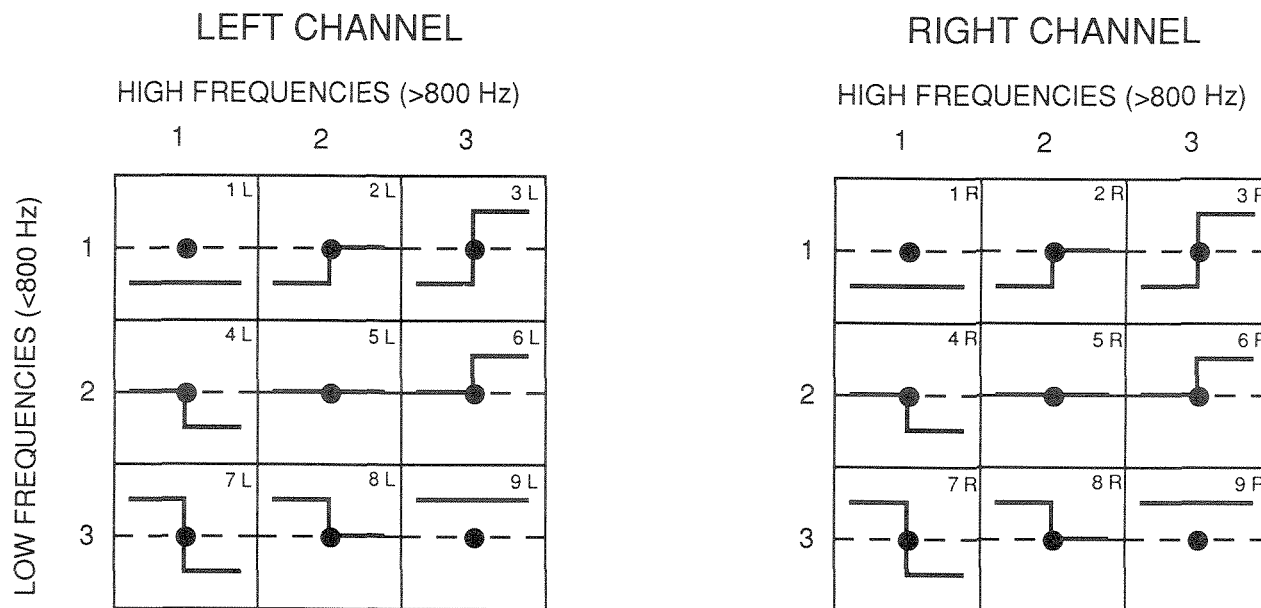
### Instrumentation

A block diagram of the binaural master hearing aid as used in the experiment is shown in **Figure 6**. As noted earlier, the binaural master hearing aid comprised two channels, one to each ear. A 2-channel tape recorder (Ampex 440C) was used as the signal source. The output of the first channel, containing speech, was led to a mixer, while the output of the second channel, containing noise, was led to a custom-built phase shifter that allowed control of interaural phase differences between the

two channels of the binaural hearing aid. The output of the two channels of the hearing aid were led to button-type hearing aid receivers (Danavox, Type SM-W) mounted in custom-made shell earmolds.

### Subjects

Subjects were eight adults between the ages of 17 and 62. All subjects had bilateral symmetrical sensorineural hearing loss. Three subjects had essentially flat audiometric configurations (<20 dB difference in threshold from 250 to 4000 Hz), and five



**Figure 5.**  
**Binaural Adjustment Matrices.** Separate matrices are shown for the left and right channels. The rows identify the loudness levels used for the low-frequency band, the columns identify the loudness levels used for the high-frequency band. Loudness levels were: Level 1 = midpoint of "soft" range, as measured using Pascoe's 1978 (28) procedure; Level 3 = upper boundary of comfort range; Level 2 = average of Levels 1 and 3.

subjects had steeply sloping audiometric configurations ( $>45$  dB difference in threshold from 250 to 4000 Hz). Word recognition scores obtained under headphones with the CID W-22 PB lists (quiet) ranged from 38 percent to 98 percent. All subjects were experienced hearing aid users who had used acoustic amplification for 1 year or longer. Four subjects were monaural hearing aid users and four were binaural hearing aid users.

### Monaural Stage

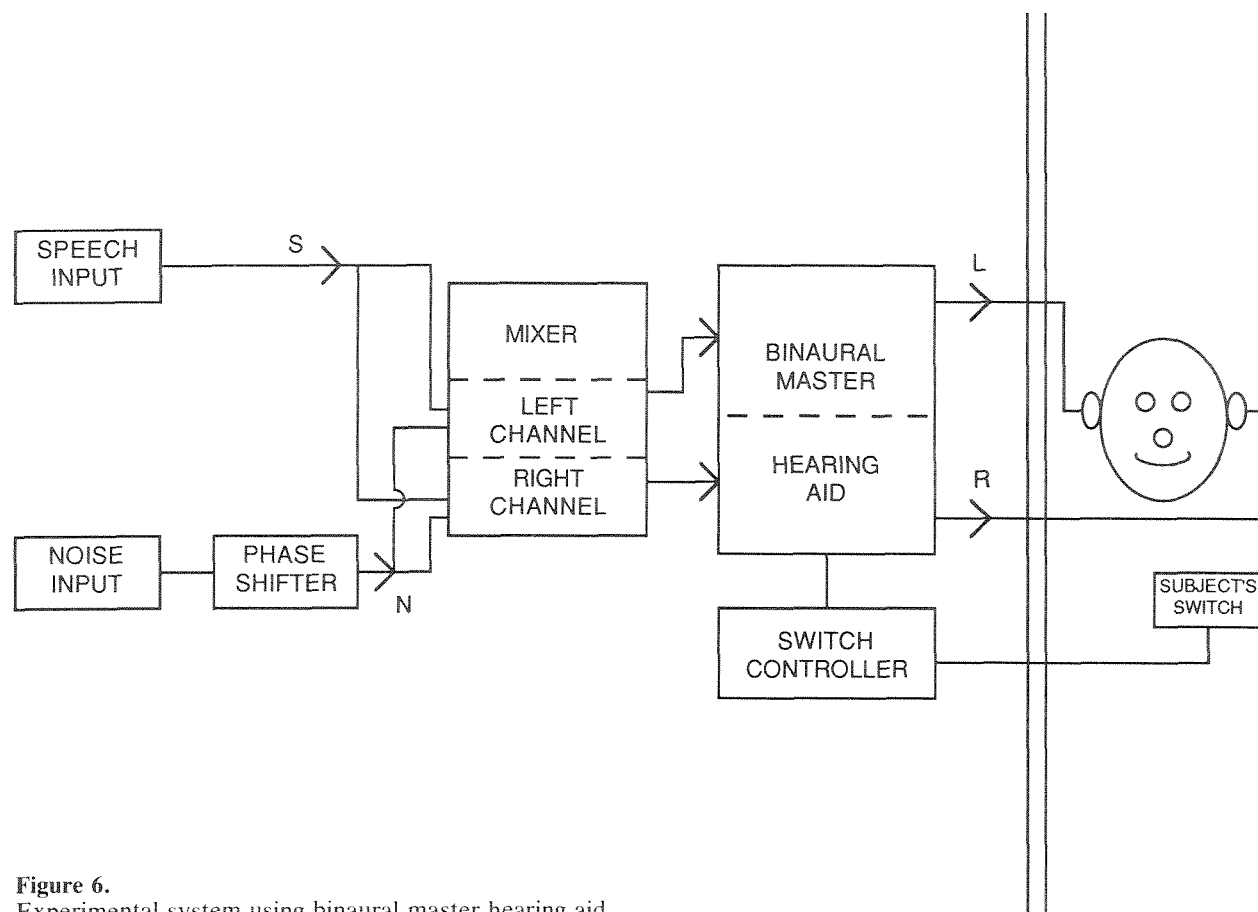
Nine frequency-gain characteristics were evaluated separately in each ear. Combinations of these frequency-gain characteristics were then evaluated binaurally.

The nine frequency-gain characteristics were based on suprathreshold measurements obtained on each subject. Loudness discomfort levels (LDL) were obtained in each ear for pure tones at the octave frequencies 250 through 4000 Hz. As before, the initial estimate of the frequency-gain characteristic was determined so that the long-term average speech spectrum was parallel to the subject's LDL curve. The dynamic range of each ear was delineated using a modification of the procedures described by Pascoe

(28). The subjects listened to a recording of continuous speech passed through either the high-pass or low-pass LDL-shaped frequency band. Three points within the subject's dynamic range were identified: the loudness levels corresponding to the upper boundary of the comfort range; the level corresponding to the midpoint of the soft range; and, the average of these two levels (this level typically fell within the comfort range). A  $3 \times 3$  matrix of frequency-gain characteristics was then formed covering all permutations of these three loudness levels in the low- and high-frequency bands.

Subjects were asked to listen to the continuous discourse passage mixed with noise ( $S/N = +15$  dB) and make paired-comparison judgments of relative intelligibility. All possible pairs of the nine frequency-gain characteristics were compared in a randomly selected sequence. The nine frequency-gain characteristics were then ranked from one to nine, based on the number of times a frequency-gain response was selected as the "winner" in each comparison. The frequency-gain characteristic most often selected as more intelligible in the paired-comparisons was ranked as first.

A comparison with speech discrimination test



**Figure 6.**  
Experimental system using binaural master hearing aid.

scores was attempted again, but the CUNY Nonsense Syllable Test was administered only to the highest, middle, and lowest ranked frequency-gain characteristics (#1, #5, and #9). It was hoped that better consistency would be obtained between paired-comparison judgments and speech-discrimination tests, since the variability introduced by allowing the subject to adjust output level was eliminated.

### Binaural Stage

The frequency-gain characteristics for the binaural phase of the experiment were selected from the monaural matrices. Two binaural matrices were evaluated. The first matrix, called the "Ranked Extremes" matrix, consisted of all permutations of those frequency-gain characteristics ranked as first, fifth, and ninth in each ear for the monaural condition. This was done with both signal and noise binaurally in phase (SoNo). The second matrix was like the first, except that the noise was reversed in phase (SoN $\pi$ ). This matrix was evaluated twice as a measure of test-retest repeatability. The third

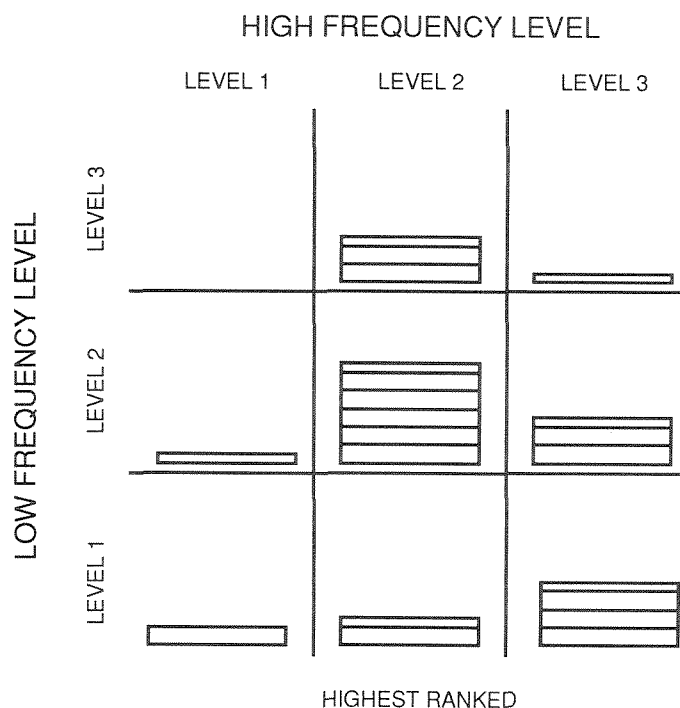
matrix, called the "Acoustic Extremes" matrix, consisted of combinations of frequency-gain characteristics that represented maximum differences between high-frequency and low-frequency amplification between ears, as well as frequency-gain characteristics that were similar at the two ears. This third matrix was included in order to evaluate combinations of frequency-gain characteristics that would not necessarily be chosen for monaural fittings, but might maximize speech intelligibility when used binaurally (9,29).

Each binaural matrix consisted of nine binaural hearing aid conditions. Again, a paired-comparison strategy was used. Subjects heard all possible pairs of the nine binaural conditions resulting in a total of 36 pairwise comparisons. The same continuous discourse passage presented in noise was used as the speech stimulus.

Speech recognition scores, using the CUNY Nonsense Syllable Test, were obtained for the binaural condition ranked as best in both the "Ranked Extremes" and "Acoustic Extremes" matrices.

**Figure 7.**

Histograms for the highest, middle, and lowest ranked frequency-gain characteristics for 16 ears. The height of the bar in each cell is proportional to the number of times the frequency-gain characteristic corresponding to that cell was chosen. The left histogram shows the choice of cells (among the 16 ears) for the highest ranked frequency-gain characteristic (Rank 1). The middle histogram shows the choice of cells for the middle ranked (Rank 5) frequency-gain characteristic. The right histogram shows the choice of cells for the lowest ranked (Rank 9) frequency-gain characteristic.



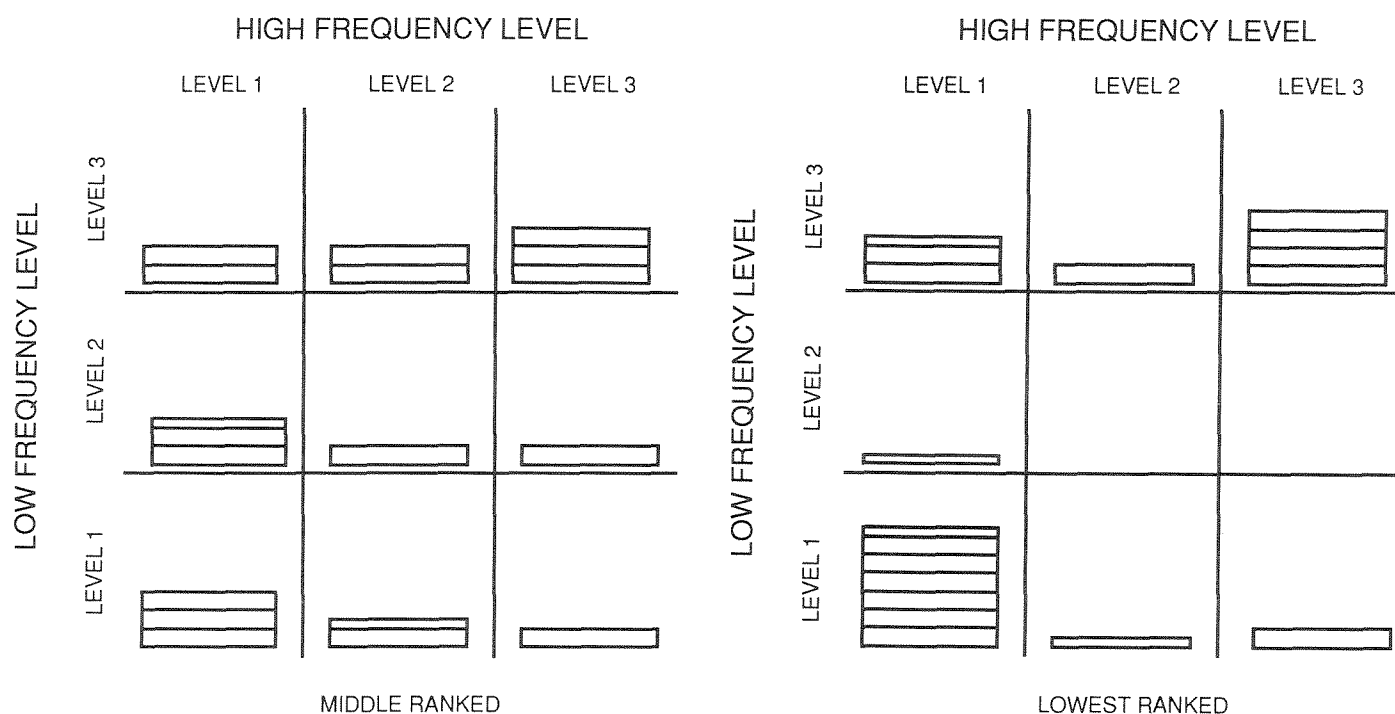
### Results: Monaural Stage

The monaural data were analyzed to determine whether or not subjects consistently chose cells other than Cell (2,2) (the first estimate of the optimum frequency-gain characteristic) as the frequency-gain characteristic through which speech was most intelligible. Recall that the frequency-gain characteristic of Cell (2,2) placed the low and high frequency bands at the subject's average comfort level and was used as the first estimate of the optimum frequency-gain characteristic. (Note that the frequency response within the low- and high-frequency bands was shaped so that the long-term average spectrum of the speech signal was parallel to the subject's LDL.) On the average, Cell (2,2) was ranked in a higher position than the other cells. For six out of sixteen ears, Cell (2,2) was ranked first. In those cases where Cell (2,2) was not ranked as best, Cell (2,2) or Cell (3,3)—the cell containing a frequency-gain characteristics that was similar in shape to Cell (2,2)—either tied with the best or was ranked in second position.

In Figure 7, histograms in which the frequency-gain characteristics ranked as first, fifth, and ninth for 16 ears are represented proportionally. The higher the bar shown in each cell, the more fre-

quently was the frequency-gain characteristic corresponding to that cell chosen. An examination of the diagram shows a movement away from Cell (2,2) as the intelligibility rank was lowered (i.e., from Rank 1 to Rank 9). In the Rank-1 condition, Cell (2,2) was chosen more frequently than any other cell. In the Rank-9 condition, Cell (2,2) was never chosen. After Cell (2,2), the second most frequently selected cell was Cell (3,3)—the frequency-gain characteristic that places low and high frequencies at the upper boundary of the subject's comfortable loudness level. The frequency-gain responses that predominate for the Rank-5 histogram are those in which either the high or low frequencies are at soft loudness levels.

The pattern is even more striking for the Rank-9 histogram. Two cells predominate; Cell (1,3), in which the loudness level is soft in the low frequencies and is at the upper boundary of the comfortable loudness range in the high frequencies; and, Cell (3,1), in which the loudness level is at the upper boundary of the comfort level in the low frequencies and at a soft level in the high frequencies. In other words, frequency-gain responses in which low and high frequencies are not balanced in loudness level are ranked poorer than those in which loudness



levels are balanced across frequencies. This supports the recommendation of Pascoe (28) and Skinner et al. (32).

A comparison of scores on the CUNY Nonsense Syllable Test for the frequency-gain characteristics ranked as first, fifth, and ninth showed a higher degree of correspondence between speech discrimination scores and paired-comparison judgments. The speech discrimination scores for 13 out of 16 cases followed the same pattern as the paired-comparison judgments. In some cases, the differences between the speech discrimination scores were small and not statistically significant.

### Results: Binaural Stage

The data were analyzed to determine whether or not the combination of the frequency-gain characteristics ranked first for the monaural condition would also be ranked as the best binaural fitting.

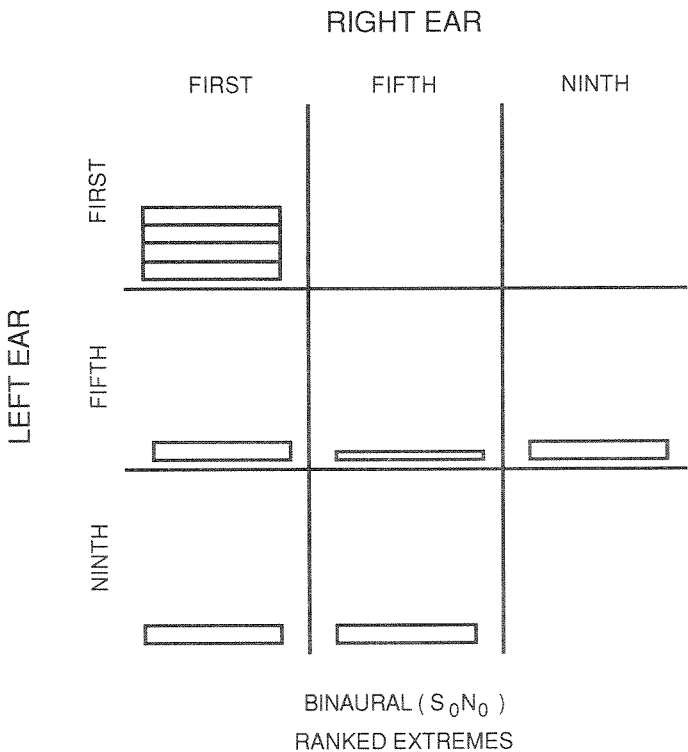
In the SoNo condition (Ranked Extremes matrix), the combination of the best monaural frequency-gain characteristics was either ranked first or tied for first for six of the eight subjects. In the remaining two subjects, the best binaural contour contained the best monaural contour in one ear. In the SoN $\pi$

condition, the combination of the best monaural frequency-gain characteristics was either ranked first or tied for first for five of the eight subjects. These results indicate that combining the highest ranking monaural conditions resulted in the highest ranked, or close to the highest ranked, binaural condition and that the ranking of the best frequency-gain characteristics was relatively unaffected by the phase of the noise in the signal. In Figure 8, histograms in which the pairs of frequency-gain responses ranked as first in the SoNo and SoN $\pi$  conditions are represented.

The speech discrimination scores for the highest-ranked binaural frequency-gain characteristic were compared to those obtained for each of the monaural frequency-gain characteristics comprising the binaural pair. The monaural and binaural scores are shown in Table 3. For all but one subject, the binaural score was equal to or higher than the better monaural score. For three of the eight subjects, the binaural score was significantly higher than the better monaural score. For only one subject was the better monaural score higher than the binaural score.

The data for the Acoustic Extremes matrix are summarized on the right side of Figure 8. The adjustment to the frequency-gain characteristic for

**Figure 8.** Histograms for the highest ranked binaural frequency-gain characteristics. For each matrix, the rows identify the frequency-gain characteristic presented to the left ear and the columns identify the frequency-gain characteristic presented to the right ear. The height of the bar in each cell is proportional to the number of times that cell was ranked as highest among the eight subjects. The histogram on the left is for the Ranked Extremes matrix with both signal and noise binaurally in phase (SoNo). The identifiers *first*, *fifth* and *ninth* indicate the rank of the frequency-gain characteristic when presented monaurally. The middle histogram is for the Ranked Extremes matrix with the signal binaurally in phase and the noise binaurally out of phase (SoN $\pi$ ). The histogram on the right is for the Acoustic Extremes matrix. In this case, the monaural frequency-gain characteristics are identified by the numbers in parentheses which specify the loudness levels of the low and high-frequency bands, respectively, (see caption to Figure 5).



each ear are identified by the numbers in parentheses shown for each row and column. For example, the identifier shown for the top row of the matrix is (1,3) indicating that the frequency-gain characteristic for the left ear corresponds to Cell (1,3) of Figure 5, (i.e., low frequencies are presented at a soft level while high frequencies are presented at the upper bound of the comfort level). Similarly, the middle row of the matrix indicating that the frequency-gain characteristic for the left ear corresponds to cell (2,

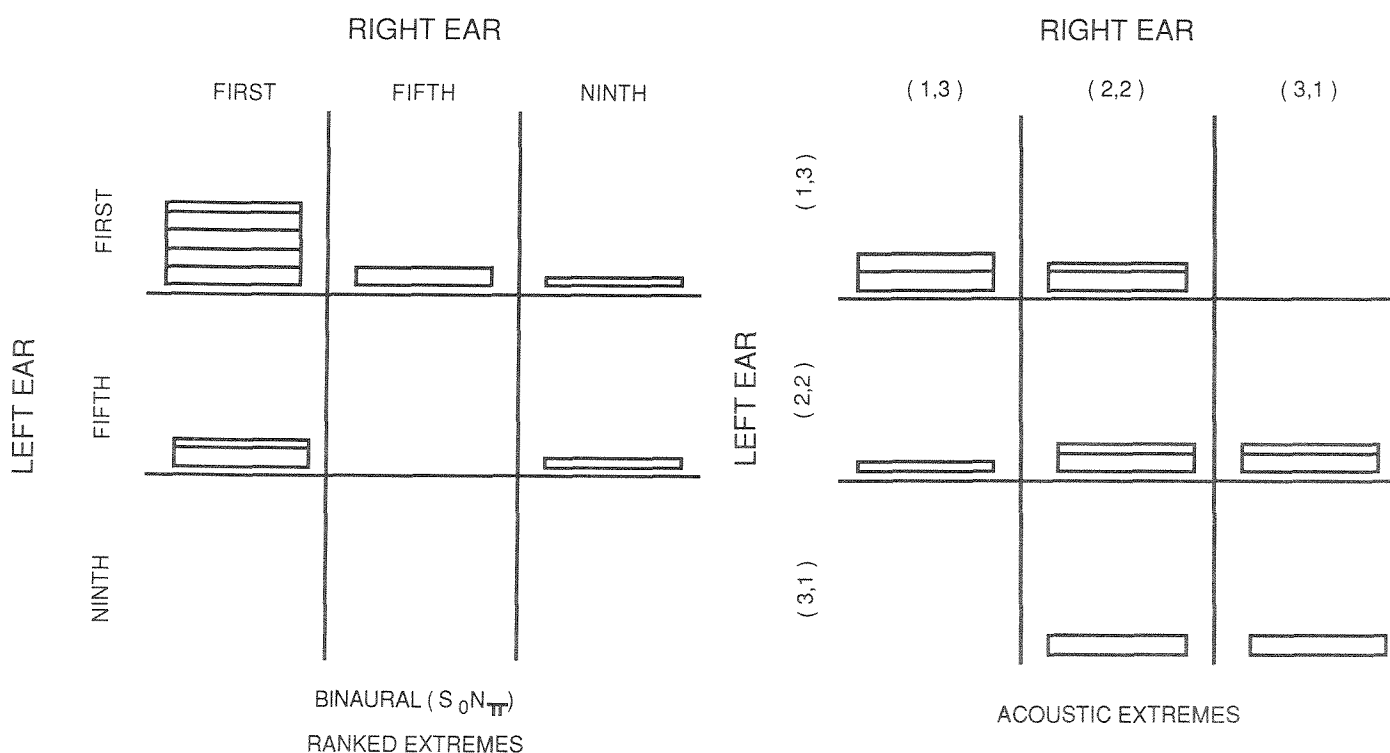
**Table 3.** Speech-discrimination scores (percent correct) for monaural condition and two binaural conditions, Ranked Extremes (SoNo) and Acoustic Extremes.

Subject	Left Ear	Right Ear	Binaural Ranked Extremes	Binaural Acoustic Extremes
1	58.2	70.0	87.3	91.8
2	68.2	56.4	70.0	65.5
3	62.7	43.6	61.8	70.0
4	71.8	28.2	67.3	71.8
5	31.8	43.6	51.8	58.2
6	73.6	70.0	73.6	81.8
7	75.5	76.4	85.5	82.7
8	71.8	88.2	76.4	76.4

2) of Figure 5. The bottom row indicates that the frequency-gain characteristic (again for the left ear) corresponds to Cell (3, 1) of Figure 5. The columns of the matrix identify the corresponding frequency-gain characteristics for the right ear, Cells (1, 3), (2, 2) and (3, 1) of Figure 5, respectively. Note that the cells lying along the diagonal of the Acoustic Extremes matrix have similar frequency-gain characteristic in each ear (allowing for differences in loudness judgements and LDL-frequency shaping between ears). The bottom left and upper right cells correspond to the most extreme binaural differences, i.e., when low frequencies are at a high intensity level in one ear they are attenuated in the opposite ear, and vice versa for high frequency components.

The histograms for the Acoustic Extremes matrix (Figure 8) show that cells along the diagonal were ranked highest, cells just off the diagonal were ranked highest slightly less often, and that the cells furthest from the diagonal were never ranked highly. These data indicate that the highest ranked binaural conditions were those for which the monaural frequency-gain characteristics were reasonably similar. Those extreme dichotic conditions in which energy in different frequency bands is boosted in opposite ears were ranked lowest of all. As shown in Table





3, the speech discrimination scores for the binaural hearing aids receiving the highest rankings in the Ranked Extremes matrices were not significantly different from the speech discrimination scores obtained for highest ranked hearing aids in the Acoustic Extremes matrix.

### General Comments

The time taken for each round-robin tournament using the paired-comparison technique was found to be on the order of 20 minutes. Thus, it would take at least 2 hours to fit a binaural hearing aid using this technique. Roughly 1 hour would be needed to obtain basic audiological data on thresholds, comfort levels, and loudness discomfort levels, and at least 1 hour of paired-comparison testing would be required (one round-robin tournament for each ear and a third tournament to check the binaural prescription). If difficulties are encountered, or if there is reason to believe that an improved estimate could be obtained with additional testing, the testing time could exceed two hours. Allowing time for breaks in testing, the total time required could become excessive for practical implementation in a clinic.

It is possible to improve the speed of testing by

use of efficient adaptive procedures. The round-robin tournament is a very effective technique for comparing all the hearing aids within a given set. It is, however, a relatively inefficient procedure if only the best hearing aids within the set are of interest. The use of a tournament strategy, such as the double elimination tournament used by Studebaker (33), would greatly improve the efficiency of the fitting procedure.

The implementation of adaptive paired-comparison strategies of the above type would require instrumentation somewhat more sophisticated than that shown in Figure 6. One approach would be to replace the switch controller shown in the diagram with a small laboratory computer. A much more powerful system would be one in which the entire system is replaced by a digital computer. The latter approach was adopted, as discussed shortly.

### Conclusions

1. For most, but not all, of the hearing-impaired subjects tested, the estimated optimum binaural condition was obtained by estimating the optimum frequency-gain characteristic independently for each ear. All of the subjects had symmetrical binaural losses.

2. The ranking of the binaural frequency-gain conditions, as obtained by the paired-comparison technique, was essentially the same for both homogenous (SoNo) and heterogenous (SoN $\pi$ ) binaural phase conditions.

3. Binaural filtering approximating the split-band technique (9,29) was ranked consistently below filtering conditions containing roughly the same frequency emphasis in both ears.

4. The majority of subjects selected a frequency-gain characteristic that amplified the speech spectrum to equal loudness levels in the high- and low-frequency bands, while paralleling loudness discomfort levels within these two frequency bands.

5. Reasonably good consistency was obtained between the paired-comparison rankings and speech discrimination scores. This was presumably because the variability was reduced by presetting loudness levels rather than allowing subjects to adjust output levels.

6. The paired-comparison procedure was found to be very efficient, but further improvements in efficiency were still needed for a practical binaural prescriptive fitting procedure. The use of a more efficient adaptive paired-comparison strategy would be useful in this regard.

#### EXPERIMENT 4: Computer-Based Hearing Aid Selection

##### Background

An early attempt to computerize the hearing aid prescription process was made concurrently with the experiments using the monaural programmable master hearing aid. The thrust of the experiment was to determine whether a computer could be of value in the prescriptive fitting of commercially available hearing aids. The prescriptive fitting protocol used was modelled on the procedures then in use with the monaural master hearing aid, except that, instead of adjusting a master hearing aid, a set of conventional hearing aids with characteristics approximating those required by the fitting protocol were used. All measurements were obtained from commercially available instruments, one being the hearing aid that was finally prescribed. A computer-search procedure was used to find a subset of hearing aids, from a much larger set of available instruments, that approximated most closely the range of char-

acteristics required by the prescriptive fitting protocol.

As before, an initial estimate of the optimum hearing aid was derived from audiological measurements obtained using a reference hearing aid. A good quality hearing aid with a reasonably flat frequency response, as measured in a standard coupler, was used as the reference hearing aid. All measurements were then referenced to the sound levels generated in the standard coupler using the reference hearing aid.

A key assumption in the above approach is that the overall gain of a given hearing aid, from sound-field to eardrum, is approximately equal to the overall gain of the reference hearing aid (which is measured on the subject) plus the difference in coupler calibrations between the given aid and the reference aid; i.e., it is assumed that

$$G_i(f) = G_r(f) + [C_i(f) - C_r(f)]$$

where  $G_i(f)$  is the overall frequency response of hearing aid  $i$ .

$G_r(f)$  is the overall frequency response of the reference aid.

$C_i(f)$  is the frequency response of hearing aid  $i$  as measured on a standard coupler.

$C_r(f)$  is the frequency response of the reference aid as measured on a standard coupler.

It is recognized that, because the acoustic impedance of the ear differs from that of a standard coupler, the above relationship is only approximate. However, it is believed that the above approximation will be reasonably good if the transducers of the reference aid (the receiver, in particular) have essentially the same characteristics as those of the aid being approximated. Since transducers of similar design are often used on different hearing aids, this is not too stringent a requirement. Another reason for expecting the above approximation to be reasonably good in practice is that it is not the change in frequency response produced by the differences in acoustic impedance (between the coupler and real-ear conditions) that is of concern, but the differential change between the reference hearing aid and hearing aid  $i$ . If the reference aid is representative of the aids being fitted, this differential change should be small.

Of the four quantities shown in the above equa-

tion,  $G_i(f)$  was the one needed for the computer-search procedure.  $G_r(f)$  was obtained from the one set of measurements involving the subject directly;  $C_i(f)$  and  $C_r(f)$  were obtained beforehand and did not involve the subject. The coupler calibrations  $C_i(f)$  represented the bulk of the measurements required for this procedure. These measurements should normally be available in any good hearing aid dispensing facility.

### Procedure

The specifics of the procedure were as follows.  $C_i(f)$  and  $C_r(f)$  were measured using standard hearing aid calibration procedures. The values of  $C_i(f)$  and  $C_r(f)$  at the standard one-third-octave band-center frequencies were then entered into a small computer. Measurements of maximum peak-to-valley deviation in the frequency response, percent harmonic distortion, and saturation sound pressure levels were also entered into the computer. All of these measures were obtained well before the subject was scheduled to be tested.

A major practical constraint was the amount of time available for testing the subject. Measurements on the subject were limited to a basic audiological workup, followed by the measurement of loudness discomfort levels (LDL) for one-third-octave bands of speech over the range 250 to 5000 Hz, plus a measurement of most comfortable level (MCL) for broadband speech. These measurements were obtained with the subject wearing the reference hearing aid.

The LDL and MCL measurements were entered into the computer, which then searched for the hearing aid that approximated most closely the frequency-gain characteristic for amplifying the long-term average speech spectrum to lie uniformly below the subject's LDL curve. By adjusting the gain control on the hearing aid, the subject was able to adjust the speech level to lie just below the LDL level over the audible frequency range, thereby placing as much of the speech signal as possible into the subject's region of residual hearing. This particular method of estimating the optimum frequency-gain characteristic was chosen because of the relatively good results obtained with it in a earlier study (1). The selection procedure used here is not limited to any specific method of individualized frequency-gain shaping and, in principle, any reasonable rule can be used for obtaining an initial

estimate of the prescribed frequency-gain characteristic.

A least-squares criterion was used in finding the closest approximation to the prescribed frequency-gain characteristic. Specifically,

$$D_i = \sum [P(f_j) - G_i(f_j)]^2$$

where  $P(f_j)$  is the prescribed frequency-gain characteristic. The summation was taken over  $j$ ,  $f_j$  being the center frequencies of the one-third-octave band filters.

The computer program then listed those hearing aids with the smallest values of  $D_i$ . Also listed were the values of maximum peak-to-valley ripple in the frequency response, percent harmonic distortion, and saturation sound pressure level. The hearing aid with the lowest  $D_i$  was chosen, subject to the constraints that the saturation sound pressure level was not excessive for the subject or that harmonic distortion or ripple in the frequency response was not unusually large.

In addition to selecting the hearing aid that provided the best least-squares approximation to the initial estimate of the optimum frequency-gain characteristic, several other hearing aids were selected to form a balanced set of hearing aids that deviated systematically from the initial estimate of the optimum hearing aid.

The original plan called for a set of four hearing aids differing in average slope from the initially prescribed frequency-gain characteristic by +6 dB/octave and -6 dB/octave above and below 800 Hz, respectively. In subsequent testing, it was found that the search procedure using the least-squares criterion for  $D_i$  defined above yielded only one or two hearing aids that met this criterion reliably. Consequently, the criteria were loosened to that of obtaining only two other hearing aids that differed from the initially prescribed frequency-gain characteristic by +6 dB/octave and -6 dB/octave, respectively, over most of the audio-frequency range. Since the initial estimate was of primary interest, two hearing aids approximating this estimate, HA1 and HA2, were selected.

The procedure was evaluated on eight subjects, all of whom were veterans with moderate to severe sensorineural hearing losses. The CUNY Nonsense Syllable Test (21) was administered for each of the selected hearing aids. Testing was done in an audiometric test suite, the subject facing the loud-

speaker delivering the test signal. Background cafeteria noise, at a 15 dB signal-to-noise ratio, was delivered via a loudspeaker facing the ear opposite the aided ear. The subject adjusted the volume control of the hearing aid to the most comfortable level prior to testing.

## Results

The results of the study are summarized in **Table 4**. The column labelled "LDL shaping" identifies the two hearing aids approximating the initial estimate of the optimum frequency-gain characteristic. Of these two hearing aids, HA1 showed the smaller least-squares deviation,  $D_i$ , from the prescribed frequency-gain characteristic.

The computer-search procedure did not always find a hearing aid in the available set of hearing aids that met the requirements of the fitting protocol. The most common problem was that a hearing aid showing a minimum least-squares deviation,  $D_i$ , for either LDL shaping +6 dB/octave or LDL shaping -6 dB/octave, turned out to be the same hearing aid that had been previously identified as minimizing  $D_i$  for LDL shaping (i.e., either hearing aid HA1 or HA2). Five of the entries in **Table 4** showed this problem. The cause of the problem was that, although the characteristics of a large number of commercially available hearing aids were stored in the data base, many of these hearing aids had similar frequency-gain characteristics, and very few had the upward sloping high-frequency characteristic typically required by the LDL-shaping rule.

Two of the subjects, S3 and S4, showed significant differences among their speech discrimination scores for the various hearing aids considered, including differences between the two hearing aids having

nominally the same frequency-gain characteristic (HA1 and HA2). Two other subjects, S5 and S6, showed significant differences in test score between HA1 and HA2, but not between the higher scoring LDL-shaping condition and the other two forms of frequency shaping. These apparently inconsistent results were obtained prior to the completion of the monaural master hearing aid study (before it was realized that these speech-discrimination scores were not sufficiently sensitive for differentiating between changes in the shape of the frequency-gain characteristic). In retrospect, it is believed that the large differences in speech-discrimination score obtained in the study were due to factors other than the shape of the frequency-gain characteristic, such as non-linear distortion, uncontrolled ripple in the frequency-gain characteristic, and internal noise.

## Conclusions

This first attempt at a computer-based prescriptive fitting protocol was deemed a failure because:

1) The range of frequency-gain characteristics required by the fitting protocol could not be obtained from the hearing aids available at that time. Even when two separate hearing aids met the criterion of minimizing  $D_i$  for two different frequency-shaping conditions, the actual differences between the two frequency-gain characteristics were small. The problem was essentially that very few commercially available hearing aids provided a substantial high-frequency emphasis, as required by the LDL-shaping rule (or similar rules for hearing aid prescription). A possible reason may be that a high-frequency boost is likely to produce acoustic feedback if the earmold does not provide an extremely good acoustic seal.

2) There were large, uncontrolled variations among the hearing aids used in the study. These variations included differences in the type and amount of distortion produced by each hearing aid, as well as differences in internal noise and related factors. In retrospect, these variations are believed to have had an effect greater than that of the controlled experimental variables.

3) The test instrument used (a conventional speech discrimination test) was insufficiently sensitive to differences between hearing aids. This method of testing was also found to be time-consuming and too inefficient for practical use in an adaptive prescriptive fitting protocol.

**Table 4.**

Speech discrimination scores for commercially available hearing aids approximating prescribed frequency-gain characteristics.

Subject	LDL Shaping		LDL Shaping + 6dB/octave	LDL Shaping - 6dB/octave
	HA1	HA2		
S1	24	none found	31	none found
S2	48	55	55	45
S3	24	3	same as HA2	52
S4	16	76	60	same as HA1
S5	53	68	68	61
S6	0	18	same as HA1	11
S7	40	39	44	same as HA1
S8	45	45	53	same as HA2

## EXPERIMENT 5: Adaptive Paired-Comparison Testing with a Digital Master Hearing Aid

### Background

The experiments reported in the preceding sections used either a programmable analog master hearing aid, or a set of computer-selected conventional hearing aids. Those experiments highlighted several serious shortcomings in conventional approaches to hearing aid prescription. The most serious problems appeared to be the low sensitivity and relative inefficiency of speech discrimination testing, the importance of variables that are not usually considered in prescriptive fitting of hearing aids, and the unknown effects of factors such as nonlinear distortion, ripple in the frequency-gain characteristic, and internal noise generated by the hearing aid. Perhaps the most serious underlying difficulty is the multivariate nature of the problem and our lack of understanding of which variables need to be controlled in order to prescribe a hearing aid for maximum benefit. The magnitude of this problem is substantially greater for binaural hearing aids than it is for monaural hearing aids.

A very useful positive result to emerge from the preceding experiments was the high efficiency and great potential value of the paired-comparison technique in the prescriptive fitting of hearing aids. Even with the use of this technique, efficiency of testing remains a problem for multivariate prescriptive fitting. The binaural hearing aid experiment highlighted both the value of the paired-comparison technique in a multivariate problem and the need to further improve its efficiency for problems of this type. The use of highly efficient adaptive paired-comparison procedures is particularly appealing from this perspective, but fairly sophisticated instrumentation is needed for implementing multivariate adaptive procedures of this type.

One approach to the implementation of adaptive paired-comparison testing is to use standard equipment under computer control. A variation of this procedure is to have the computer control a combination of conventional analog amplifiers and digital filters. Several leading research groups have adopted the latter approach (M. Haggard, personal communication with the authors, 22,34).

An alternative approach is to simulate the hearing aids of interest on a computer. Although the value of computer simulation as a research tool was

recognized some time ago (13), it was not until the development of high-speed array processors that the technique became sufficiently practical for real-time testing of simulated hearing aids. The array processor became generally available at about the time the decision was made to computerize the programmable master hearing aid used in the preceding experiments. Consequently, it was decided to use real-time computer simulation rather than computer-control of conventional analog equipment. The underlying rationale was that computer simulation represented a more general approach, and allowed the use of advanced signal-processing techniques in addition to implementing multivariate adaptive paired-comparison procedures.

### Instrumentation

Figure 9 shows a block diagram of the digital master hearing aid. Note that two computers are used. The first is the array processor (MAP-300), the second is the controlling computer (DEC LSI-11/23). The input to the system can be either acoustical or electrical. For an acoustical input, the signal is first converted into electrical form by a microphone and then amplified by a low-noise pre-amplifier. The electrical input is hard-wired and requires signals at typical line-voltage levels, such as those produced by a tape recorder/reproducer. A third possible input is that of an FM receiver. This form of input is used when simulating a wearable hearing aid. In this case, the microphone is mounted in a conventional hearing-aid case (either a behind-the-ear or an in-the-ear unit can be used). The output of the microphone is amplified and delivered to an FM transmitter worn by the user. The output of the FM transmitter is picked up by an FM receiver at the input to the computer system. The output of the computer system is then transmitted back to the user by means of a second FM link operating at a different carrier frequency.

The input to the computer system is first low-pass filtered to the operating bandwidth of the hearing aid so as to avoid aliasing errors. The input signal is then digitized using a 12-bit analog-to-digital (A-D) converter. The sampling rate of the A-D converter is under the control of the array processor and can be as high as 40 kHz. In order to simulate a hearing aid with more than one input (e.g., a binaural hearing aid), a multiplexer is included in the A-D converter unit. The digital output of the A-

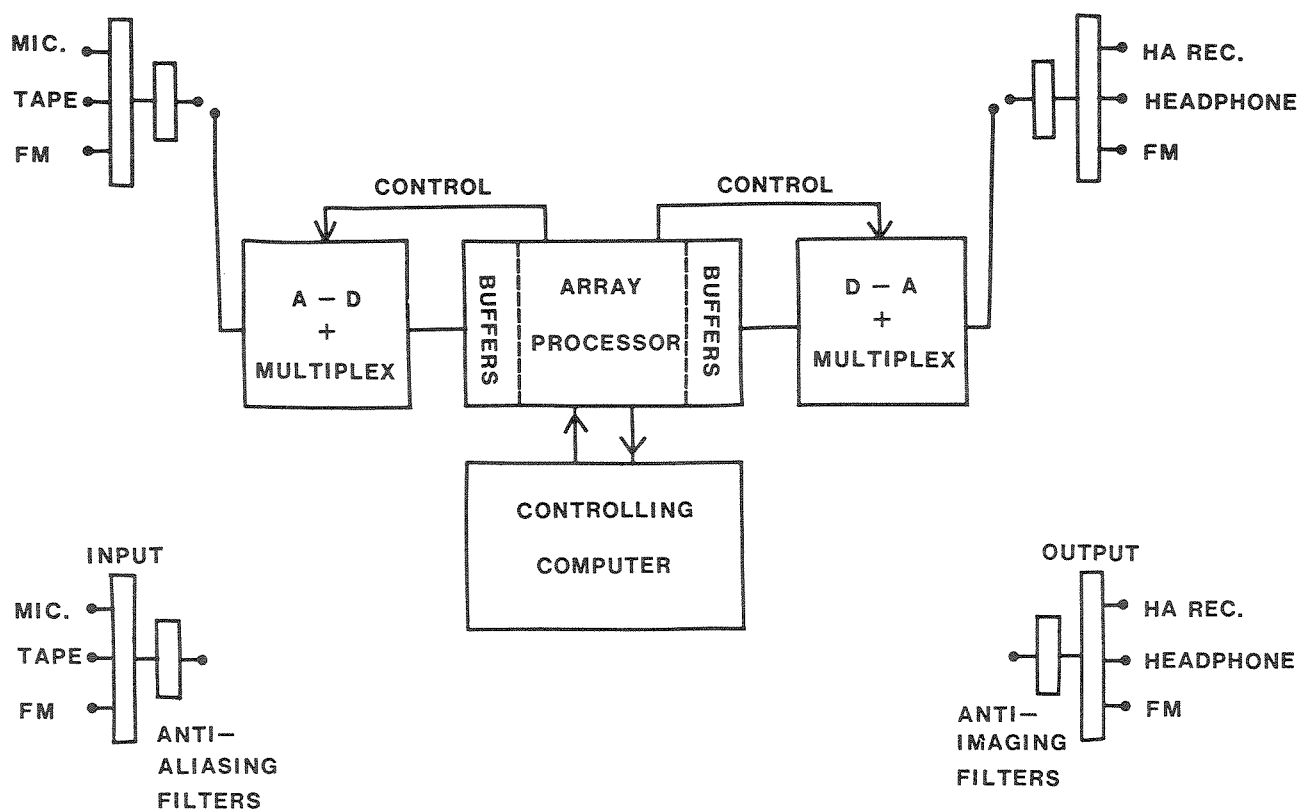


Figure 9.  
Block diagram of digital master hearing aid.

D converter is operated on by the array processor according to the instructions received from the controlling computer. The output of the array processor is converted back to analog form by the digital-to-analog (D-A) converter, demultiplexed if necessary, passed through a low-pass anti-imaging filter to eliminate spurious frequency components outside the bandwidth of the hearing aid, and then fed to an appropriate transducer. The latter may be a conventional hearing aid receiver, or audiometric earphones, or similar transducers. If an FM link is used in simulating a wearable hearing aid, then the output of the anti-imaging filter is fed to the computer-based FM transmitter.

The configuration described above is similar in concept to that of a computer-controlled digital filter except that an array processor is used instead of a digital filter. The simpler arrangement of using a computer-controlled filter is much favored in the development of wearable digital hearing aids. The advantage of using an array processor is that highly sophisticated signal-processing techniques can be implemented in addition to digital filtering. The array

processor is particularly well suited for performing fast Fourier transforms in real time, thereby allowing the signals to be processed directly in terms of their short-term frequency spectra.

The controlling computer not only instructs the array processor as to which signal-processing operations should be performed at any given time, but it can also be programmed to implement specific prescriptive fitting strategies as well as controlling experiments and analyzing data. In the experiment described below, the array processor was programmed not only to simulate a master digital hearing aid, but also to simulate an automated audiometer used in obtaining the basic audiological data required for the prescriptive fitting strategies to be evaluated.

### Experimental Procedure

The purpose of the experiment was to compare three adaptive paired-comparison procedures for use in the prescriptive fitting of hearing aids. The first of these procedures, the round-robin technique, is one that could be implemented without a computer

(as demonstrated in the experiments reported above). The second procedure, the double-elimination tournament, can be implemented using instrumentation of moderate complexity. The third method, the simplex procedure (11), is statistically more efficient than the other two procedures, but requires a computer for rapid selection and control of the experimental conditions needed by this form of adaptive paired-comparison testing. The procedure was modified slightly to allow for direct comparisons with the estimates obtained with the other two adaptive strategies (25).

Eight adult subjects with sensorineural hearing impairments ranging from moderate to severe participated in the experiment. In the first stage of the experiment the computer system was programmed to serve as an audiometer in order to obtain auditory thresholds, comfort and discomfort levels, as well as various loudness levels as a function of frequency. A procedure based on that developed by Pascoe (28) was used. As in the previous experiments, these data were obtained for the reference hearing aid condition.

A first estimate of the optimum frequency-gain characteristic was obtained by computing the gain necessary to place the long-term average speech spectrum at a comfortable level at all frequencies within the bandwidth of the hearing aid (5 kHz). The frequency-gain characteristic was then adjusted adaptively using each of the three above mentioned adaptive strategies to determine if an improved estimate of the frequency-gain characteristic could be obtained. Details of the adaptive procedures and methods of testing are provided by Neuman et al. (25).

### Results and Conclusions

All three adaptive procedures converged on essentially the same frequency-gain characteristic for each subject. This was a very reassuring result because it indicated that not only were the adaptive procedures consistent with each other but also that the response surface had a single major peak. Each of the adaptive strategies was designed to converge on the peak of the response surface, the coordinates of this peak being the parameter values defining the optimum frequency-gain characteristic. As in the previous experiments, the variables adjusted were the gains above and below 800Hz. A major concern was that the response surface might have more than

one major peak, in which case different estimates of the optimum frequency-gain characteristic would be obtained depending on the initial estimate. This appears not to be the case.

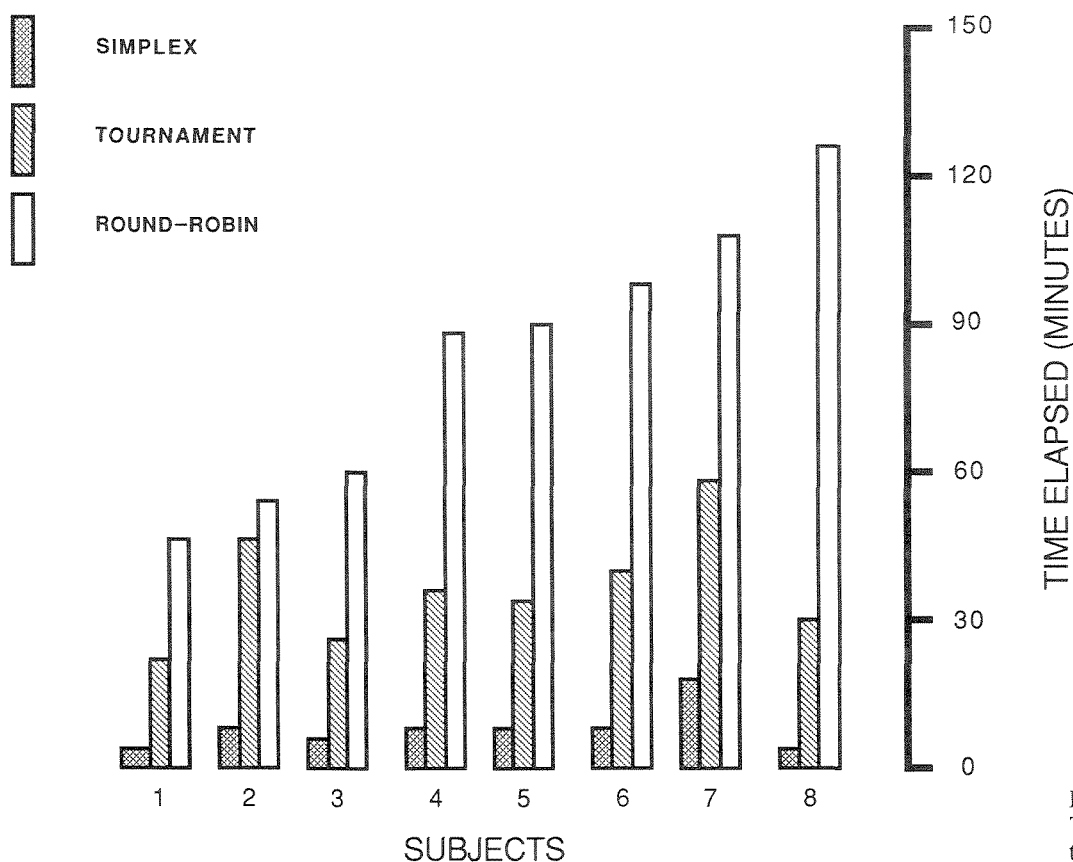
An important caveat is that different estimates of the optimum frequency-gain characteristic might be obtained for different test material. All of the testing in this experiment involved paired-comparison judgments of continuous speech by a male talker against a background of cafeteria noise. The optimum frequency-gain characteristic may differ for a female or child's voice, or for different types of background noise. Sullivan et al. (36) have already shown that relative preferences (in paired-comparison testing) for different frequency-gain characteristics will vary as a function of signal level.

It is also important to bear in mind that the frequency-gain characteristic was subdivided into only two frequency regions during the adjustment procedure. A more precise estimate of the optimum frequency-gain characteristic will be obtained if the frequency range is divided into smaller subdivisions. The cost of doing this is an increased number of variables to be adjusted, which, in turn, will add significantly to the time taken to converge on the optimum frequency-gain characteristic.

Large between-subject differences were observed among the estimated optimum frequency-gain characteristics. For only one subject was the final estimate equal to the initial estimate. For three of the eight subjects, the final estimate differed from the initial estimate primarily in the amount of overall gain and only slightly in the shape of the frequency response. For the remaining four subjects, the final estimate of the optimum frequency-gain characteristic differed significantly in both shape and overall gain from the initial estimate. These results emphasize the importance of individualized prescriptive fitting of hearing aids.

A question of primary interest in this study was the time taken by each of the adaptive strategies in converging on the estimated optimum setting. The results of the study showed that, as expected, the round-robin procedure was the least efficient and the simplex technique the most efficient of the three procedures. The relative increase in efficiency in going from the round-robin to the tournament to the simplicial strategies, however, was greater than anticipated. **Figure 10** shows that the average time taken for the simplex procedure was a fraction of





**Figure 10.**  
Time taken by each subject on three adaptive strategies.

that required by the double-elimination tournament and an order of magnitude less than that required by the round-robin technique.

The average time taken by the simplex procedure to converge on the estimated optimum frequency-gain characteristic was on the order of 10 minutes. This time period is short enough to be of practical use in a clinical setting, particularly if estimates of the optimum frequency-gain characteristics are to be obtained for different conditions of hearing aid use (e.g., for different speakers and various types of background noise).

### CONCLUDING COMMENTARY

The series of experiments described above trace the development of adaptive paired-comparison testing as a technique for the prescriptive fitting of hearing aids. The equipment needed to implement the technique became progressively more complicated as the range of applications of the paired-

comparison technique were explored. In the first few experiments, the limitations of the equipment imposed serious practical constraints on the experimental procedures and, by implication, on the implementation of the prescriptive fitting protocol in a practical clinical setting. The use of a digital master hearing aid altered the situation dramatically. As illustrated by the last experiment, the inherent limitations in applying adaptive paired-comparison testing are no longer technological, but are of a more fundamental nature. How many frequency bands should be used in the adjustment procedure? Which hearing aid should be prescribed if the estimated optimum frequency-gain characteristic differs substantially between conditions of hearing aid use? These are very basic questions that require serious thinking about our whole approach to hearing aid design, evaluation, and prescription.

The computational power of the digital master hearing aid used in the last experiment was considerably more powerful than required for that form of testing. Simpler digital hearing aids have since been



developed consisting primarily of computer-controlled digital filters (22,34). Recent advances in high-speed digital-signal-processing (DSP) chips have opened up the possibility of a practical, wearable digital hearing aid being developed in the near future (5,8,27).

The array-processor digital hearing aid developed for Experiment 5 is currently being used to investigate advanced forms of signal processing, such as the Weiss-Aschkenazy method of noise reduction (37) and the orthogonal-polynomial method of multiband amplitude compression (20). The results of these experiments have not shown statistically significant improvements in speech intelligibility but, in both cases, the quality of the processed signals was judged to be superior to the original, unprocessed signals.

The development of practical digital hearing aids for personal use should provide a solution to the problem encountered in Experiment 4. In this experiment it was found that the frequency-gain characteristics available in the current generation of conventional hearing aids provided only a crude approximation to the estimated optimum frequency-gain characteristic. There are also many practical considerations that digital hearing aids can address, but which have hitherto been neglected because of their complexity. These include difficulties encountered in the acoustic coupling between hearing aid receiver and eardrum, amplifier noise, and both linear and nonlinear distortions introduced by the hearing aid.

The introduction of digital hearing aids presents new forms of acoustic distortion that need to be considered in the design, evaluation, and prescriptive fitting of hearing aids. In addition to the familiar forms of nonlinear distortion in analog hearing aids—such as clipping and other nonlinear operations, and various forms of harmonic and intermodulation distortion—digital hearing aids can also introduce distortion produced by the digitization process, such as quantization noise and the possibility of aliasing errors. See the tutorial review in this issue (13) for a description of these effects.

Experiments are currently in progress investigating the various forms of distortion, both linear and nonlinear, produced by a hearing aid. In one experiment, the effects of unwanted ripple in the frequency response of a hearing aid was studied by simulating the ripple typically produced by the

acoustic tubing of a behind-the-ear hearing aid (35). The results showed that a peak-to-valley ratio of 3 dB for normal-hearing listeners and a 6 dB ratio for hearing-impaired listeners could be tolerated before any change in the quality of an amplified speech signal was detected. In another ongoing study, a general index for specifying distortion in hearing aids is being developed (18).

Another basic issue of some concern is that the judgmental criteria used in evaluating modern signal-processing hearing aids may differ significantly from those used for conventional hearing aids. For example, Experiment 3 showed a relatively good correspondence between hearing aids ranked by paired-comparison judgments and by speech discrimination testing. Similarly, Sullivan et al. (36) showed a good correspondence between paired-comparison judgments of relative intelligibility and relative quality. All of the above studies, however, were concerned with the shape of the frequency-gain characteristic in a conventional hearing aid.

The relatively good correspondence between paired-comparison judgments of intelligibility and quality may not hold for hearing aids involving nonconventional forms of signal processing. Data obtained in an ongoing experiment using an experimental amplitude compression system have shown that improvements in sound quality can be obtained at the expense of a small reduction in intelligibility, and *vice versa*. Similar results also have been obtained with various forms of signal processing to reduce background noise (26). In general, these experiments indicate that, with more advanced forms of signal processing, important compromises may need to be made between the conflicting demands of improved intelligibility and improved sound quality.

In conclusion, the experiments reported in this paper illustrate how necessity, being the mother of invention, lead to the development of a digital master hearing aid. Once weaned, however, the instrument has shown a wide variety of new applications not originally conceived of by its parent.

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## REFERENCES

1. **Abramovitz R:** Frequency shaping and multiband compression in hearing aids. *J Acoust Soc Am* 65, Suppl. 1, S136 (A), 1979.
2. **Collins MJ and Levitt H:** Comparison of methods for predicting optimum functional gain. In *Acoustical Factors Affecting Hearing Aid Performance*, GA Studebaker, I Hochberg (Eds.), 341–354. Baltimore: Univer. Park Press, 1980.
3. **Collins MJ and Levitt H:** Evaluation of a protocol for fitting a wearable master hearing aid. *Bul Prosthet Res*, 18(1):35–40, 1981.
4. **Cox R**, personal communication.
5. **Cummins K and Hecox K:** Ambulatory testing of a wearable digital hearing aid. In *Proceedings of the 10th Annual Conference on Rehabilitation Technology*, R D Steel and W Gerrey (Eds.) Washington, D.C.: RESNA, 1987.
6. **Dubno JR and Dirks DD:** Evaluation of hearing-impaired listeners using a nonsense-syllable test: 1. Test reliability. *J Speech Hear Res* 25:135–141, 1982.
7. **Edwards AL:** *An Introduction to Linear Regression and Correlation*. San Francisco: W. H. Freeman and Company.
8. **Engebretson AM, Morley R, and Popelka GA:** A unified digital hearing-aid design and fitting procedure. *ASHA*, 25:101 (Abstract), 1983. See also Engebretson et al., *J Rehabil Res Dev* 24(4), 1987 (this issue).
9. **Franklin B:** The effect on consonant discrimination of combining a low-frequency passband in one ear and a high-frequency passband in the other ear. *J Audit Res* 9:365–378, 1970.
10. **Hertzano T, Levitt H, and Slosberg R:** Computer-assisted hearing aid selection. *Speech Communication Papers, 97th Meeting of the Acoustical Society of America*, New York, 1979. See also *J Acoust Soc Am* 65 Suppl 1, S137(A), 1979.
11. **Levitt H:** Adaptive testing in audiology. In *Sensorineural Hearing Impairment and Hearing Aids*, C Ludvigsen and J Barfod (Eds.) *Scand Audiol Suppl* 6, 1978.
12. **Levitt H:** A cancellation technique for the amplitude and phase calibration of hearing aids and nonconventional transducers. *J Rehabil Res Dev* 24(4), 1987 (this issue).
13. **Levitt H:** Digital Hearing Aids: A tutorial review. *J Rehabil Res Dev* 24(4), 1987 (this issue).
14. **Levitt H:** Functional Gain Measurements of Amplitude and Phase. In *Proceedings of the 10th Annual Conference on Rehabilitation Technology*, RD Steel and W Gerrey (Eds.), 416–417. Washington, D.C.: RESNA, 1987.
15. **Levitt H:** Hearing impairment and sensory aids: A tutorial review. *J Rehabil Res Dev* 23(1):xiii–xviii, 1986.
16. **Levitt H and Collins MJ:** An experimental protocol for the prescriptive fitting of a wearable master hearing aid. In: *Acoustical Factors Affecting Hearing Aid Performance*, 323–339, Baltimore: University Park Press, 1980.
17. **Levitt H, Collins MJ, Dubno JR, Resnick SB, and White REC:** *Development of a Protocol for the Prescriptive Fitting of a Wearable Master Hearing Aid*. Communications Sciences Laboratory Report No. 11, Center for Research in Speech and Hearing Sciences, City University of New York, 1978.
18. **Levitt H, Cudahy E, Jiang WH, Kennedy E, and Link C:** Towards a general measure of distortion. *J Rehabil Res Dev* 24(4), 1987 (this issue).
19. **Levitt H, Neuman A, Mills R, and Schwander T:** A digital master hearing aid. *J Rehabil Res Dev* 23(1):79–87, 1986.
20. **Levitt H, Neuman A., and Toraskar J:** Orthogonal polynomial compression amplification for the hearing impaired. In *Proceedings of the 10th Annual Conference on Rehabilitation Technology*, RD Steel and W Gerrey (Eds.), 410–412. Washington, D.C.: RESNA, 1987.
21. **Levitt H and Resnick SB:** Speech reception by the hearing impaired: Methods of testing and the development of new tests. In *Sensorineural Hearing Impairment and Hearing Aids*, Ludvigsen CV and Bartod J (Eds.), *Scand Audiol Suppl.* 6, 105–130, 1978.
22. **Levitt H, Sullivan J, and Hwang J:** The chameleon: A computerized hearing aid measurement/simulation system. *Hearing Instruments* 37 (2):16–18, 1986.
23. **Levitt H, White REC, and Resnick SB:** Prescriptive fitting of wearable master hearing aids: A progress report. In *Hearing Aids: Current Developments and Concepts*, M Rubin (Ed.), 149–159. Baltimore: Univ. Park Press, 1976.
24. **Morgan DE, Wilson RH, and Dirks, DD:** Loudness discomfort level: Selected methods and stimuli. *J Acoust Soc Am* 56: 577–587, 1974.
25. **Neuman A, Levitt H, Mills R, and Schwander T:** An experimental comparison between three adaptive hearing-aid fitting strategies. *J Acoust Soc Am*, in press.
26. **Neuman A, Levitt H, Weiss M, and Schwander T:** Evaluation of noise reduction techniques for hearing aids. In *Proceedings of the 10th Annual Conference on Rehabilitation Technology*, RD Steel and W Gerrey (Eds.), 413–415. Washington, D.C.: RESNA, 1987.
27. **Nunley J, Staab W, Steadman J, Wechsler P, and Spencer B:** A wearable digital hearing aid. *Hear J* 29–31, 34–35, October, 1983.
28. **Pascoe DP:** An approach to hearing-aid selection. *Hearing Instruments* 29:12–16, 36, 1978. See also Pascoe DP: Frequency response of hearing aids and their effects on the speech perception of hearing-impaired subjects. *Ann Otol Rhinol Laryngol*, Suppl. 23, 84, 1975.
29. **Rosenthal RD, Lang JK, and Levitt H:** Speech reception with low-frequency speech energy. *J Acoust Soc Am* 57:949–955, 1975.
30. **Shore I, Bilger RC, and Hirsh IJ:** Hearing-aid evaluation: Reliability of repeated measurements. *J Speech Hear Disorders*, 25:152–170, 1960.
31. **Siegel S:** *Nonparametric Statistics for the Behavioral Sciences*. New York: McGraw Hill, 1956.
32. **Skinner MW, Pascoe DP, Miller JD, and Popelka GK:** Measurement to determine the optimal placement of speech energy within the listener's auditory area: A basis for selecting amplification. In *The Vanderbilt Hearing-Aid Report*, GA Studebaker, FH Bess (Eds.), Monographs in Contemporary Audiology, 161–169. Upper Darby, PA, 1982.
33. **Studebaker GA:** Hearing aid selection: An overview. In *The Vanderbilt Hearing-Aid Report*, GA Studebaker and FH Bess (Eds.), 147–155. Upper Darby, PA, 1982.
34. **Studebaker GA, Sherbecoe RL, and Matesich JS:** Spectrum shaping with a hardware digital filter. *J Rehabil Res Dev* 24(4), 1987 (this issue).
35. **Sullivan J, Levitt H, Hwang J, and Hennessey AM:** Discriminability of frequency-response irregularities in hearing aids. In *Proceedings of the 10th Annual Conference on Rehabilitation Technology*, RD Steel and W Gerrey (Eds.), 407–409. Washington, D.C.: RESNA, 1987.
36. **Sullivan JR, Levitt H, Hwang J, and Hennessey AM:** An experimental comparison of four hearing-aid selection procedures. Submitted for publication, *Ear and Hearing*.
37. **Weiss MM and Aschkenasy E:** Wideband speech enhancement. *RADC-TR-81-53 Final Technical Report*, May 1981.